Biomechanical Determinants of Injury Risk and Performance during Change of Direction: Implications for Screening and Intervention

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ABBREVIATIONS

2D: Two-dimensional
3D: Three-dimensional
ACL: Anterior cruciate ligament
ADFA: Ankle dorsi-flexion angle
ADFM: Ankle dorsi-flexion moment
AUC: Area under curve
BM: Body mass
BW: Body weight
COD: Change of direction
CODD: Change of direction deficit
COF: Cut-off frequency
COM: Centre of mass
COP: Centre of pressure
CG: Control group
CMAS: Cutting movement assessment score
CV%: Coefficient of variation
CMC: Correlation of multiple coefficients
CI: Confidence interval
CUT45: 45° cutting task
CUT90: 90° cutting task
DVJ: Drop vertical jump
EMG: Electromyography
ES: Effect size
F-MARC 11+: FIFA's Medical Assessment and Research Centre 11+
FFC: Final foot contact
FPPA: Frontal plane projection angle
GCT: Ground contact time
GRF: Ground reaction force
HBF: Horizontal braking force
HFA: Hip flexion angle
HFM: Hip flexion moment
HRA: Hip rotation angle
HPF: Horizontal propulsive force
HGRF: Horizontal GRF
IC: Initial contact
ICC: Intraclass correlation coefficient
IFPA: Initial foot progression angle
IG: Intervention group
KAA: Knee abduction angle
KAM: Knee abduction moment
KFA: Knee flexion angle
KFM: Knee flexion moment
KIRM: Knee internal rotation moment
LOA: Limits of agreement
LESS: Landing error scoring system
ML: Medio-lateral
MLGRF: Medio-lateral GRF
NSCA: National Strength and Conditioning Association
PEA: Percent exact agreement
PFC: Penultimate foot contact
QASLS: Qualitative analysis of single leg loading
QTM: Qualisys Track Manager
RMANOVA: Repeated measures Analysis of variance
ROC: Receiver operating characteristics
ROM: Range of motion
SD: Standard deviation
SDD: Smallest detectable difference
SEM: Standard error of mean
SET: Sequential estimation technique
SPM: Statistical parametric mapping
SSC: Stretch shortening cycle
SWC: Smallest worthwhile change
TE: Typical error
TJA: Tuck jump assessment
TS: Trial size
VPF: Vertical propulsive force
VBF: Vertical braking force
VGRF: Vertical GRF
WA: Weight acceptance
XOC: Crossover cut

ABSTRACT

Changing direction is a key action linked to successful performance in multidirectional sports, but also an action associated with non-contact anterior cruciate ligament (ACL) injuries. Despite the importance of directional changes for sports performance and its association with ACL injury risk, biomechanical studies generally investigate performance and injury risk determinants independently. Preliminary evidence indicates that the mechanics and techniques associated with faster change of direction (COD) performance are at odds with safer (i.e., reduced knee joint loads) COD, known as the "performance-injury conflict". Therefore, understanding the mechanics and techniques, screening tools, and training interventions that can reduce the relative risk of injury during COD actions, while improving performance, are of great interest to practitioners working with multidirectional athletes.

The primary aims of the thesis are three-fold: 1) to identify the biomechanical determinants of performance and injury risk during cutting, in order to better understand the potential "performance-injury conflict"; 2) to validate a qualitative screening tool against three-dimensional (3D) motion analysis for evaluating cutting movement quality and identifying athletes who display high peak knee abduction moments (KAM); and 3) to understand the biomechanical effects of a COD technique modification intervention, with the aim of reducing knee joint loads while maintaining or enhancing cutting performance.

Initially, Study 1 (Chapter 4) aimed to examine the effect of trial size on the within- and between-session reliability measures and outcome values for cutting biomechanics. Cutting joint angle, ground reaction force (GRF), and knee flexion moment (KFM) variables demonstrated high reliability, while knee internal rotation moments (KIRMs), KAMs and angles demonstrated high variability and bias between sessions. Increasing trial size had a negligible effect on reliability measures and outcome values; thus, trial sizes of three to five appear adequate to capture reliable cutting biomechanical data, with no apparent requirement for greater trial sizes.

Study 2 (Chapter 5) aimed to investigate the biomechanical determinants of cutting performance and injury risk (knee joint loads). The primary findings were that techniques and mechanics associated with faster cutting performance (i.e., faster velocities over the COD, greater final foot contact [FFC] braking forces over shorter ground contact times (GCT), greater FFC KFMs, smaller hip and knee flexion angles and range of motion (ROM), wider lateral foot plant distances, and greater internally rotated foot postures) are in direct conflict with safer cutting mechanics (i.e., reduced knee joint loading), and support the concept of a "performance-injury conflict".

Study 3 (Chapter 6) aimed to validate a qualitative screening tool for identifying "high-risk" cutting mechanics and peak KAMs, against 3D motion analysis. The screening tool, titled the "Cutting Movement Assessment Score" (CMAS), was a 9-item tool, based on a literature review and findings from Study 2 containing the "high-risk" postures and mechanics. A strong relationship was observed between CMAS and peak KAMs, and trials and participants who displayed greater CMASs typically displayed "higher-risk" postures and greater multiplanar knee joint loads.

Finally, studies 4 and 5 (Chapters 7 & 8) aimed to examine the effects of a COD technique modification intervention on cutting performance and injury risk biomechanics. Study 4 served as a feasibility study which was performed in an applied setting. The primary findings were that 6-weeks COD technique modification training with externally directed verbal coaching cues improved cutting performance and movement quality (i.e., reduced CMASs) in male youth soccer players.

Study 5 expanded on the feasibility study by comprehensively monitoring the biomechanical changes (3D motion and GRF analysis) in response to COD technique modification. The primary findings were that 6-weeks COD technique modification training with externally directed verbal coaching cues resulted in significant and meaningful improvements in cutting performance. When examining group means, no statistically significant or meaningful changes in multiplanar knee joint loads were observed; however, there was considerable individual variation in response to the training intervention, with athletes considered "higher-risk" responding positively to the intervention.

Overall, these studies in combination help to provide further insight into the "performance-injury conflict" that is present during cutting, and overall assist in the development of more effective COD speed and ACL injury mitigation training programmes. The findings from the thesis provide practitioners with: 1) a greater insight into the coaching of faster and safer COD; 2) a field-based qualitative screening tool for evaluating cutting movement quality; and 3) training interventions that can be easily implemented in the field with minimal equipment to improve COD performance and potentially reduce knee joint loading.

Keywords: anterior cruciate ligament; injury risk profiling; movement screening; cutting; side-step

CHAPTER 1: INTRODUCTION

1.1 Introduction

Change of direction (COD) is defined as a reorientation and change in the path of travel of the whole-body centre of mass (COM) towards a new intended direction (118, 621), and the ability to rapidly change direction is an important action associated with successful performance in multidirectional sports (51, 171, 283, 409, 487, 510, 549, 628). COD manoeuvres (e.g., side-step, crossover cuts [XOC], split-steps, and pivots) are frequently performed in sports such as soccer (51, 487), netball (184, 549), and rugby (499, 601, 632), with soccer players performing ~600 cuts of 0-90° and ~100 turns of 90-180° during games (51), while directional changes of 45°, 90°, and 180° have recently been established as frequently performed actions in netball (549). Furthermore, Fox et al. (184) quantified the attacking and agility actions performed in netball, finding the side-step was the most commonly performed agility technique (on average across positions) followed by the shuffle, split-step, stop-and-back, and spin manoeuvre. Importantly, COD manoeuvres are linked to decisive actions in these sports, such as evading an opponent to penetrate the defensive line in rugby (tackle-break success in rugby) (402, 601, 632), or getting into space to receive a pass in netball (184), while COD actions are also linked to goal scoring and assists in soccer (171). As such, due to the importance of COD in multidirectional sports, and providing the physiological and mechanical basis underpinning agility (412) (unplanned movements in response to stimuli), practitioners are interested in technical guidelines and training strategies that enhance COD performance.

Changing direction has also been identified as an action associated with several injuries, such as lateral ankle sprains (178) and groin strains, and development of athletic groin pain (191). However, of concern is that changing direction is a key action associated with non-contact ACL injuries in numerous multidirectional sports (soccer, rugby, handball, netball, Australian rules football, American football, and badminton) (52, 63, 94, 170, 270, 291, 298, 376, 426, 583), due to the propensity to generate high multiplanar knee joint loading (flexion, rotation, and abduction moments) during plant foot contact (39, 126, 127, 273, 301), thus increasing ACL strain (26, 288, 342, 424, 513). Although ankle and groin injuries occur more frequently than non-contact ACL injuries in multidirectional sports (390, 617) and undoubtedly are of concern to practitioners, ACL injuries are a debilitating and arguably more devastating and potentially career threatening injury with a plethora of short- and long-term consequences (financial, health, and psychological) (112, 241, 315, 333, 468), with an elevated and earlier risk of developing osteoarthritis a problematic issue (333, 571). An estimated 250,000 ACL injuries occur annually in the United States (241) and two million worldwide (483), with over one billion dollars, estimated, spent annually on the reconstruction and rehabilitation in the United States. ACL injuries often require surgery (210); thus, extensive rehabilitation periods are required, resulting in prolonged absence and the potential to lose sporting scholarships or contracts (375), while athletes who do successfully return to sport post ACL reconstruction, may demonstrate reduced sports related performance, reduced number of appearances, and shorter career longevity (313, 368). As such, due to the severity and potential aforementioned negative implications of non-contact ACL injury, understanding the mechanics, techniques, and training interventions that can reduce the relative risk of non-contact ACL injury during COD actions, while improving performance, are of great interest to practitioners working with multidirectional athletes, and thus, the primary focus for this thesis.

Given the importance of directional changes for sports performance and its association with ACL injury risk, it is somewhat surprising that the majority of COD biomechanical studies investigate performance (143, 275, 338, 346, 498, 530, 531, 597) and injury risk determinants (117, 127, 139, 267, 272-274, 301, 358, 516, 519, 593, 625) independently. From a performance perspective, greater braking and propulsive forces and impulses over short GCTs have been identified as mechanical determinants of faster COD speed performance (143, 228, 338, 346, 498, 530, 531, 596, 597). Additionally, whole-body kinetics and kinematics such as greater ankle power, ankle plantar-flexor moments, torso lean and rotation, hip power and extensor moments, rapid knee and hip extension, wider lateral foot plants, and low COM are also associated with faster cutting performance (228, 346, 597); highlighting the importance of the lower-limb triple extensor musculature and trunk lean towards the intended direction of travel. Conversely, from an injury perspective, COD techniques with a wide lateral foot plant (127, 228, 272, 301), greater hip abduction angles (519, 593), increased internal initial foot progression angles (IFPA) (274, 519), increased internal initial hip rotation angles (228, 358, 516, 519), greater peak and initial knee abduction angles (KAA) (272, 274, 301, 358, 516), greater lateral trunk flexion over the plant foot (127, 189, 267, 272), smaller knee flexion angles (KFA) (76, 593) and greater GRF (273, 516, 519) are associated with greater KAMs (synonymous with knee valgus moments) and thus injury risk (242, 342, 343, 512, 614). However, less is known regarding the mechanics and techniques necessary for optimal (faster) COD performance and how they relate to injury risk (183, 228).

There is evidence, although limited, which indicates the techniques and mechanics required for faster COD performance are in direct conflict with techniques and mechanics required for safer (i.e., lower knee joint loads) directional changes (115, 183, 228). For instance, COD techniques such as increased internal IFPAs and pelvic and hip internal rotation angles are associated with greater KAMs (127, 274, 519), but may be optimal for COD performance due to effective alignment of the whole-body COM into the new intended direction (118, 229, 274). Lateral trunk flexion (over the plant foot contact) has been shown to increase knee joint loading (126, 127, 272); however, this strategy may be adopted by athletes to deceive (feint) opponents (61, 62, 252). Importantly, wider lateral foot plants are also associated with greater KAMs (126, 127, 228, 272, 301), where larger moment arms and KAMs are created with a more medial whole-body position with respect to the foot, and centre of pressure (COP) positioning more lateral to the COM of the body and tibia (228, 274). However, a wide lateral foot plant is required to generate medio-lateral (ML) GRF and impulse to accelerate into the new direction (228, 261, 272). Smaller KFAs are associated with greater KAMs (76, 593) and anterior tibial shear (42, 43, 343, 344, 506, 615), but increasing knee flexion angle can prolong GCTs duration and reduce exit velocity, both of which are key factors related to faster COD performance (143, 213, 228, 338, 346, 498, 530, 531, 596, 597).

Specifically, Havens and Sigward (228) revealed faster cutting performance was associated with greater lateral foot plant distances, ML impulse, and internal hip rotation angle, though it is worth noting that greater KAMs were also observed with wider lateral foot plants, which may increase ACL injury risk. This study, to the best of the author's knowledge, is the first and only published study to investigate both COD injury risk and performance biomechanical determinants, indicating that techniques and mechanics that facilitate faster performance may also expose athletes to greater KAMs and potentially greater injury risk. Moreover, Fox (183) has recently discussed in a narrative review that the techniques and mechanics for ACL injury mitigation during COD, may not necessarily be optimal for COD performance. Collectively, these studies and biomechanical principles suggest that there is a "performance-injury conflict" during COD, which is problematic for practitioners who aim to improve their athletes' performance and reduce risk of injury. Further insight is required to confirm whether COD mechanics required for effective performance also increase knee joint loads and potential ACL injury risk.

Although ACL injuries are a complex interaction of internal and external risk factors (hormonal, anatomical, biomechanical, neuromuscular, environmental) (48, 241, 365, 469), there are modifiable risk factors (biomechanical and neuromuscular control) which practitioners can target and address to potentially mitigate ACL injury. Mechanically, ACL injuries occur when a load exceeds the ligaments tolerance capacity (327, 365), and multiplanar knee joint loads generated during COD plant foot contact (39, 126, 127, 273, 301) have the potential to load and strain the ACL (26, 288, 342, 424, 513). These knee joint loads are amplified during COD when aberrant and "high-risk" movement patterns are displayed (i.e., lateral trunk flexion, knee valgus etc.) (126, 127, 183, 228, 241, 272, 301); however, these biomechanical deficits are modifiable through appropriate training and conditioning (126, 127, 228, 272, 301, 432, 435), which may reduce knee joint loading during COD actions and subsequent risk of non-contact ACL injury (241, 432, 435). Subsequently, training interventions that can reduce KAMs, and multiplanar knee joint loads are viable strategies to reduce ACL injury risk (138, 183, 241, 301, 432, 435).

Reducing knee joint moment magnitudes can be achieved via reducing the moment arm (particularly in the frontal plane), GRF, or a combination of the two (301). Although there are various training modalities that can be used to alter COD biomechanics, COD technique modification specifically is an effective modality for reducing knee joint loads or factors linked to greater knee joint loads (76, 115, 126, 127, 277). Researchers have reported reductions in peak KAMs as a result of cutting technique modification via alterations in lateral foot plant distance (and orientation) and trunk alignment performed acutely (within-session) (127) and chronically over a 6-week intervention (126). Additionally, Celebrini et al. (76) also found instructing athletes to move their COM closer to their base of support (i.e., increase knee flexion and reduced lateral foot plant distance) reduced cutting KAMs. However, these studies have failed to consider the implications of such changes in technique on COD performance (i.e., completion time, GCT, and exit velocity). Bringing the plant foot closer to the midline can reduce KAMs by reducing moment arm distance, but could be suboptimal for ML impulse generation and may result in suboptimal COD performance (228, 261, 272). As athletes are driven by performance, they may be unlikely to adopt movement strategies which decrease knee injury risk if they do not result in effective performance (183, 228). As such, exploration into the effects of COD technique modification on both performance and knee joint loading is warranted to improve our knowledge of the potential "performance-injury conflict" during directional changes. Conducting such research into the relationship and interaction between performance and injury risk determinants during COD may assist in the development of more effective ACL injury mitigation, COD speed, and agility programmes (183).

Historically, the kinetics and kinematics of the FFC during directional changes have been analysed in terms of injury risk and performance (108, 126, 127, 301, 346, 358, 359, 362, 516, 519, 530, 531). However, changing direction is a multistep action (11), and emerging research has revealed that the braking characteristics exhibited during the penultimate foot contact (PFC) of a COD play an important role in deceleration prior to changing direction, particularly during CODs ≥ 60˚ (11, 143, 202, 227, 272-275, 290, 403). Researchers have shown that athletes that produce greater magnitudes and proportions of horizontal braking forces (HBF) in the PFC, relative to the FFC, subsequently experience reductions in GRFs and KAMs in the turning or cutting limb (272-274); thus, reducing knee joint loads and potential risk of injury. In addition, promising results from a performance perspective have also shown that a braking strategy which emphasises a greater proportion of HBF in the PFC, relative to the FFC, are associated with faster 180° COD performance (143, 202, 275). Therefore, it may be advantageous for athletes to apply greater magnitudes of HBF in the PFC, because braking earlier with greater magnitudes will increase braking impulse. By increasing braking impulse, based on the impulse-momentum relationship, this will lead to a greater reduction in horizontal momentum of the COM (i.e., change in momentum; net deceleration) to allow more effective weight acceptance (WA) and preparation for drive-off propulsive forces during the FFC, while concurrently reducing knee joint loads in the turning limb (147); where most ACL injuries occur. However, it is unknown whether longitudinally coaching COD techniques, which emphasise a braking strategy of greater magnitudes of HBF in the PFC are effective in reducing knee joint loads, while maintaining or producing faster COD performance, and is therefore an area of future research.

The ability to identify athletes' potentially "at-risk" (i.e., "higher-risk") of injury is a critical step in effective ACL injury risk reduction (185, 241). 3D motion and GRF analysis is considered the gold standard for evaluating movement kinetics and kinematics (187, 241); however, it is a complex, time-consuming, and expensive process, which is often restricted to laboratory settings, thus limiting its application for practitioners working in field-based settings (163, 187, 241, 302, 408, 527). As such, qualitative screening tools such as the Landing Error Scoring System (LESS) (430, 434), Tuck Jump Assessment (TJA) (238, 391, 397), and Qualitative Analysis of Single Leg Loading (QASLS) (8, 237) have been developed, offering practitioners an easier and cost-effective method to assess lower-limb and whole-body postures associated with potentially increased risk of ACL injury, typically during landing tasks. Although screening landing mechanics is indeed applicable for jump-landing sports (netball, basketball, volleyball) where the primary action associated with non-contact ACL injury is landing (54, 248, 310, 541), these aforementioned assessments may lack specificity to the unilateral, multiplanar plant-and-cut manoeuvres observed when changing direction (187, 278, 372). This is particularly important when practitioners aim to screen athletes who participate in sports such as soccer, handball, American football, badminton, and rugby where directional changes are a primary action associated with non-contact ACL injuries (52, 63, 94, 170, 270, 291, 298, 376, 426, 583). Furthermore, there are mixed findings as to whether examination of landing mechanics can identify athletes with poor cutting mechanics (4, 302, 421, 551), with evidence to suggest that an athlete's mechanics and "injury risk profile" are task dependent (279, 302, 388). As such, there is a requirement for the development and validation of a field-based qualitative screening tool to evaluate and identify poor movement quality and "high-risk" mechanics specifically for cutting.

1.2 Aims

The primary aims of the thesis were three-fold:

1) To identify the biomechanical determinants of performance and injury risk during cutting, in order to better understand the potential "performance-injury conflict" (Chapter 5).
2) To validate a qualitative screening tool against 3D motion analysis for evaluating cutting movement quality and identifying athletes who display high peak KAMs (Chapter 6).
3) To understand the biomechanical effects of a COD technique intervention, with the aim of reducing knee joint loads while maintaining or enhancing cutting performance (Chapter 7 & 8).

1.3 Research questions and rationale

The results from this body of research should help answer the following research questions:

What is the effect of trial size on the within- and between-session reliability measures and outcome values for cutting biomechanics? (Chapter 4, Study 1)

The majority of COD biomechanical investigations fail to report their reliability measures, but in order for practitioners to be confident in the data collected, and subsequent interpretation of such data, it is integral to understand the reliability and variability of the assessment (16, 27, 254), to establish "real" changes in performance (16, 254, 474). Trial size is an important methodological consideration in the design of human movement experiments because of its effect on statistical power (23) and reliability measures (257). Only one study (370) has examined the effect of trial size on cutting biomechanics reliability; however, the study only examined up to 5 trials. Some studies use up to 10 trials when investigating COD biomechanics, thus the effect of trial size on COD reliability and outcome values using trials sizes as high as 10 warrants inspection.

Do COD techniques and mechanics which result in faster performance have a concurrent increased risk of injury? (Chapter 5, Study 2)

Research regarding COD biomechanics generally investigates performance and injury risk determinants independently, with little information regarding the techniques required for faster performance and how they relate to injury risk. Havens and Sigward (228), to the best of the author's knowledge, is the only study to examine the biomechanical determinants of cutting performance and KAMs; however, they did not examine PFC braking characteristics, velocity profiles, and did not examine KIRMs. Conducting such research into the relationship and interaction between performance and injury risk determinants during COD may assist in the development of more effective ACL injury mitigation, COD speed, and agility programmes.

Do braking strategies which emphasise greater PFC braking forces help alleviate knee joint loading and facilitate faster performance during cutting? (Chapter 5, Study 2)

Emerging research has shown promising results regarding the role of PFC braking characteristics on COD performance and reducing knee joint loading; however, this research must be expanded with larger sample sizes to confirm whether PFC braking strategies are effective in facilitating faster performance and alleviating knee joint loads.

Can a qualitative screening tool identify athletes with "high-risk" movement mechanics and high peak KAMs during cutting? (Chapter 6, Study 3)

Qualitative screening tools exist for evaluating jump-landing mechanics; however, no such tool for evaluating cutting movement quality has yet to be created or validated. As such, a qualitative screening tool should provide scientists and practitioners with a cheaper, more efficient, and applicable method to identify athletes displaying "high-risk" mechanics during cutting tasks, so informed decisions can be made regarding the future training for that athlete.

What are the biomechanical effects of a six-week COD technique modification intervention on performance and injury risk biomechanics? (Chapter 7 & 8, Study 4 & 5)

Finally, COD technique modification interventions have been shown to reduce knee joint loads (126, 277); however, these interventions did not contain control groups (CG) and failed to acknowledge measurement errors, while the implications of such technique changes on cutting performance are unclear. For greater insight into COD in terms of performance and risk of injury, more detailed analysis considering both aspects are warranted. If COD technique modification is successful (i.e., reducing biomechanical characteristics associated with increased risk of injury without negatively affecting performance), practitioners could implement the technique modification programme as a potential ACL injury mitigation programme in applied field-settings.

CHAPTER 2: LITERATURE REVIEW

The content of this literature review focuses on four key areas. Firstly, ACL injuries are discussed in Chapter 2.1 with a specific focus on the mechanisms of injury. Secondly, injury risk screening is discussed in Chapter 2.2 with a primary emphasis on field-based screening tools. Thirdly, the biomechanical determinants of injury risk and performance during COD are comprehensively examined in Chapter 2.3 with additional emphasis placed on training interventions that reduce ACL loading during COD. Finally, methodological and technical issues in 3D motion and GRF data collection and analysis are discussed in Chapter 2.4.

2.1 Anterior cruciate ligament injuries

2.1.1 Anatomy

One of four ligaments in the knee, the ACL is a collagenous structure that originates from the medial wall of the lateral femoral condyle, running anteriorly, medially and distally, inserting into a fossa, anterior and lateral to the medial tibial spine (medial tibial eminence) (158, 450, 633), and consists of two fibre bundles: the anteromedial and posterolateral bundle (449). The primary function of the ACL is to restrain anterior tibial translation and the secondary function is to restrain transverse rotation (158, 633), thus playing a crucial role in knee joint stability (158, 633).

2.1.2 ACL injury implications

ACL injuries are debilitating, with negative economic (112, 241, 468), psychological (241, 315), and health (241, 333) implications for athletes and the general population. Annual ACL injury rates are estimated to be 250,000 in the United States (241) and two million injuries worldwide (483), with over one billion dollars, estimated, spent annually on ACL reconstruction and rehabilitation in the United States (468). ACL injuries typically require surgery (210); thus, extensive rehabilitation periods are required, typically resulting in the loss of a sport season, and the potential to lose sporting scholarships or contracts (375). In addition, ACL injuries have also been found to affect athletes' psychological wellbeing with elevated mood disturbances, reduced confidence, and fear of re-injury reported during rehabilitation (232, 333, 382, 588). However, Langford et al. (315) found that athletes who successfully returned to competitive sport displayed a more positive psychological response to sports participation (higher scores on ACL return to sport injury scale) compared to athletes who did not return to competitive sport). While in the long term, from a knee joint health perspective, ACL injuries can lead to joint instability (210), meniscal tears (604), and earlier and increased risk of developing osteoarthritis (333, 463, 571), yet it is not uncommon for athletes or the general population to require additional surgery for meniscus and cartilage damage following ACL reconstruction (210). Thus, both short- and long-term effects appear to be evident for those affected by this serious injury (Figure 2.1).

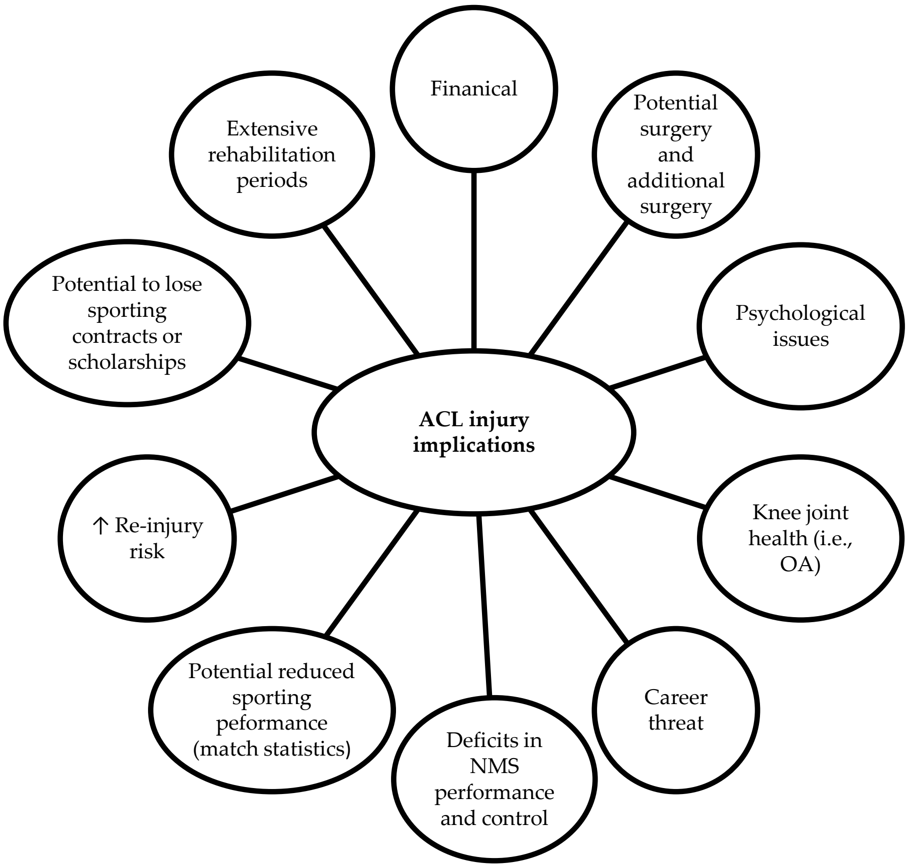


Figure 2.1. ACL injury implications (OA: Osteoarthritis; NMS: Neuromuscular)

The results of a meta-analysis revealed that only 55% of non-elite athletes (i.e., not playing professionally or at the highest level of sport) who sustained an ACL injury return to competitive level sport (14). Recently, however, a systematic review revealed a return to sport rate of 83% in elite (i.e., professional or playing at the highest level of sport) athletic populations (American professional sports), with a mean time of 6-13 months (313). Yet, these findings still indicate approximately 1 in 5 elite athletes sustain a career-ending injury (313). Additionally, of concern, ACL re-injury rates of ~29.5% are reported within 2 years return to sport (9% ipsilateral and 20.5% contralateral) (443), while long term follow up periods ≥ 10 years have observed re-injury rates of 23-27% (57, 153, 377). In particular, incidence rates of a second ACL injury following ACL reconstruction are significantly greater within the first year of reconstruction compared to the second (209, 401), especially within young active athletes (401, 605). This finding could be attributed to athletes returning to sport too early before a complete biological (ligamentisation, bones bruises, healing of joint tissue), and functional recovery of knee (neuromuscular control, strength, rate of force development, proprioception) has been attained (401), or presence of non-modifiable risk factors, or modifiable risk factors which have not been addressed.

For example, deficits and asymmetries in lower-limb strength, rate of force development, neuromuscular control, and functional performance between limbs have been reported post ACL injury (typically greater within a year of ACL reconstruction) (12, 197, 296, 316, 401, 442, 444, 492, 559). Kyritsis et al. (312) recently found athletes that did not meet a return to sport clinical discharge criteria in 6 tests (three of which required >90% limb symmetry indexes in hopping tasks, isokinetics, running T-test, and on-field sport-specific rehabilitation) were four times more likely to sustain a second ACL rupture. Moreover, Toole et al. (559) reported a greater proportion of athletes who achieved return to sport criterion cut-offs (isokinetic strength and hop symmetry indexes) maintained the same level of sports participation, compared to athletes who did not meet the cut-off following ACL reconstruction. As such, this has led to the suggestion and recommendations of delaying return to sport for two years post-ACL reconstruction to allow full biological and functional recovery of the knee (401) (and to encourage sufficient rehabilitation), so the knee can withstand the external forces and loads experienced during sport. However, having an athlete delay their return to sport for two years and missing at least two sports seasons may have direct consequences on the athlete's career or financial sustainability, and therefore may not be feasible.

Although a large proportion of athletes do return to sport following ACL injury (14), and go on to have successful careers (165, 167, 223, 313, 336, 368), it is feared that athletes may not have the capacity to perform to the same skill level or athleticism post ACL injury. Recently, two systematic reviews have explored the effect of ACL injury on sports-related performance outcomes post-reconstruction and, in general, indicate a potential reduction in sporting performance (match statistics and player efficiency ratings), playing appearances, and career longevity (313, 368). However, it should be noted that these sports related effects vary between individuals and are dependent on the sport and playing position (69, 73, 165-168, 190, 223, 285, 313, 336, 368, 520). Moreover, the majority of studies have focused on post ACL injury sports-related outcomes and performance in male athletes and American-based sports such as basketball, ice hockey, American football, Major League Soccer, and baseball (69, 73, 165-168, 190, 223, 285, 520); thus, further research is required in female athletes, athletes from different sports, and sports played in different parts of the world (i.e., Europe, Asia, etc.). It should also be acknowledged that the potential decrements in performance following ACL injury may be attributed to the progression of time, rather than the result of ACL injury/reconstruction (313), and successful return to sport will also be influenced by other factors such as an athlete's quality and intensity of rehabilitation and physiological wellbeing and motivation.

Collectively, ACL injuries can have numerous health, financial, physiological, neuromuscular, and performance implications for athletes and the general population (Figure 2.1); thus, mitigating ACL injury risk and the ability to "profile" and identify potentially "higher-risk" athletes are of fundamental importance. In order to do so, however, it is imperative to understand the aetiology and mechanisms of ACL injury, risk factors, and subsequent mitigation strategies.

2.1.3 Mechanisms of ACL injury

ACL injury risk factors are multifactorial (anatomical, hormonal, biomechanical, neuromuscular, environmental) (241, 469) and a complex interaction of both internal (within the body) and external (outside the body) factors (48, 365, 469). Figure 2.2 presents ACL injury risk factors based on recent narrative and systematic reviews (1, 393, 451, 467), which categorises these risk factors as non-modifiable, modifiable, and potentially modifiable. Non-modifiable factors include sex, genetic predisposition, family history and anatomical factors, such as intercondylar notch width and tibial slope, which can predispose athletes to greater risk of sustaining a non-contact ACL injury, while hormonal factors such as the phase of the menstrual cycle for females can result in greater ACL injury risk. Additionally, biomechanical and neuromuscular factors are also linked to ACL injury risk due to increasing knee joint loads, but importantly are modifiable. Finally, some environmental factors can also contribute to ACL injury risk and are possibly (depending on the situation) modifiable, such as footwear choice; however, athletes and practitioners cannot control weather, opponents etc.

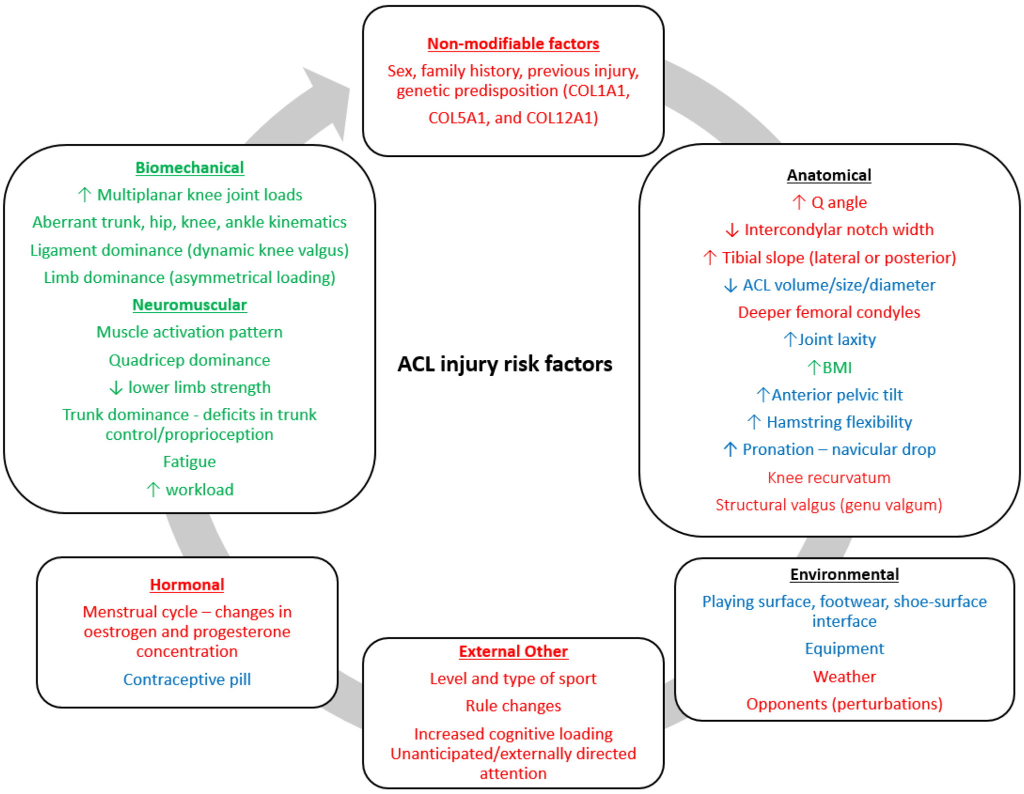


Figure 2.2. ACL injury risk factors adapted from systematic and narrative reviews (1, 393, 451, 467). Green = modifiable, red = non-modifiable, and blue = possibly modifiable (but has constraints/ requires further research).

In simplistic terms, injuries to the ACL occur when an applied load exceeds the ligament's tolerance (327, 365), and non-contact ACL injuries generally occur during high impact tasks such as landing, deceleration, and COD actions (52, 63, 94, 170, 270, 291, 298, 426, 583). Because some ACL injury risk factors are non-modifiable (Figure 2.2), such as intercondylar notch, tibial slope, and genetic predisposition, much attention has been placed on modifiable factors: biomechanics and neuromuscular control deficits (i.e., "high-risk" and aberrant movement mechanics), which can influence knee joint loads (39, 126, 127, 273, 301) and potential ACL strain (26, 288, 342, 424, 513). In order to better understand the mechanisms and risk factors of ACL injury in sport, various research approaches have been adopted (Figure 2.3).

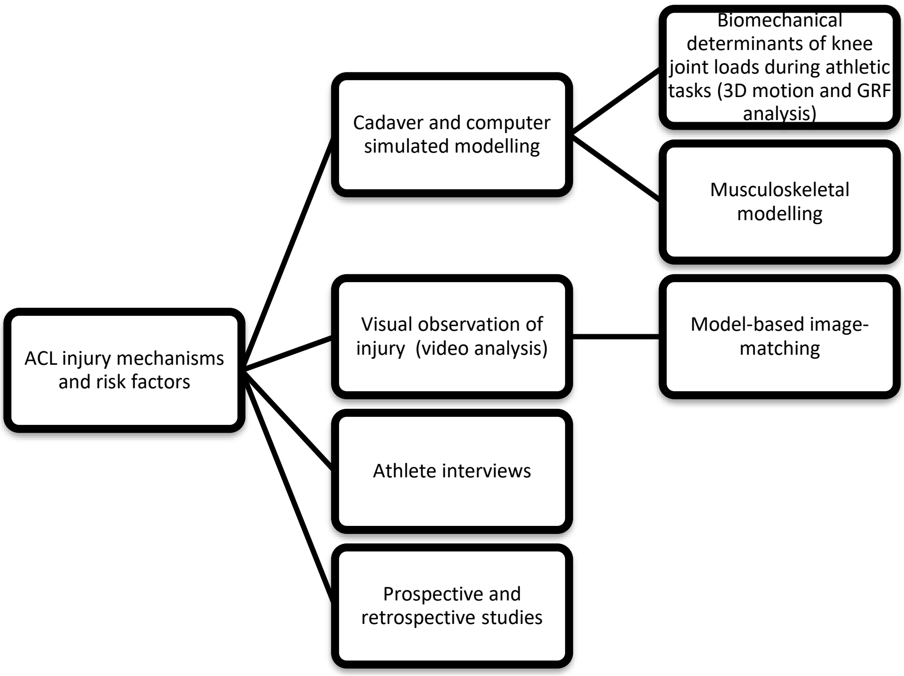


Figure 2.3. Research approaches to understand the mechanisms and ACL injury risk factors. Adapted from Krosshaug et al. (307)

Athlete interviews have been used to understand the scenarios and actions of ACL injury, but can be influenced by recall bias (307). Cadaveric and computer simulated studies have been performed in an attempt to understand the mechanism of ACL ruptures, indicating that increased quadriceps loading and anterior tibial shear at limited knee flexion angles (~0-40°) increase ACL strain (42, 43, 343, 344, 615), attributed to the patella ligament angular attachment to the tibia (31, 43, 176). However, McLean et al. (357), via computer simulated modelling, demonstrated that sagittal plane biomechanics alone cannot injure the ACL in isolation. Additionally, external knee abduction moments (valgus) (342, 343, 512, 614) and tibial internal rotation moments have been shown to increase ACL strain (342-344), yet researchers have shown that a combination of loading in several planes (combined anterior tibial shear, knee abduction and internal rotation moments) increases ACL strain to a greater extent than uniplanar loading alone (26, 288, 342, 424, 513). Additionally, external KFMs have been reported to be associated with proximal tibia anterior shear (506, 631). As such, the mechanism of ACL injury, from a knee joint loading perspective, has been stated as "multiplanar" (469), thus highlighting the importance of minimising multiplanar joint loading to mitigate ACL injury risk (138, 327, 328). Because multiplanar knee joint loads have the propensity to strain the ACL, this has led to biomechanical investigations into the kinetic and kinematic determinants of knee joint loads during athletic tasks using 3D motion and GRF analysis (Chapter 2.3.3), and using multiplanar knee joint loads as surrogates of non-contact ACL injury risk (39, 126, 242, 358). Additionally, it is worth noting that high KAMs are also associated with patellofemoral pain (PFP) (393, 395, 580) and also used as surrogates of PFP injury risk. More recently, musculoskeletal modelling of athletic tasks has been used to assess ACL loading (522, 524, 591).

Although cadaveric and computer simulated studies provide insight into the mechanism of ACL injury, it is vital to acknowledge the limitations of these investigative approaches. Often cadaveric specimens are generally older than athletic populations who sustain ACL injuries, which may contain different tissue properties, structural degradation, and possess lower bone mineral density, and these findings are generalised across knees (584). Additionally, these are simulated events which provide no information regarding the actual inciting events of ACL injuries in sport. Moreover, it is important to note that the external moments are applied to the cadaver knee joint without muscular support (126). Muscular co-contraction has the potential to support and assist in ligament unloading (328). As such, in combination with cadaveric and computer simulated research, it is important to understand the events which incited the ACL injury (kinematics, sport-specific situations, athletes' behaviour).

2.1.4 Visual, qualitative, and quantitative inspection of the inciting events of ACL injuries

It is important to note that ACL injuries occur from contact and non-contact mechanisms, which are defined as follows (94, 376):

- Contact: when an injury occurs with bodily contact (direct: to the injured limb, indirect: contact to another body part) with another player.
- Non-contact: when the injury was caused with no external contact with another player.

Visual and qualitative inspection of ACL injury video footage have been performed in a range of multidirectional sports, including soccer (63, 204, 583), handball (426), netball (541), Australian rules football (94), rugby (376), basketball (310), American football (270), and badminton (291), with non-contact ACL incident rates of 44-72% reported for team sports, and 100% for badminton. Generally, these studies indicate that non-contact ACL injuries occur during a COD (side-step, cut or turn, plant-and-cut manoeuvre) (52, 63, 94, 170, 270, 291, 298, 376, 426, 583), landing (predominantly single-leg weight bearing) (94, 291, 426, 541, 583), and deceleration actions (54, 63, 94); however, the primary action of injury will be influenced by the predominant activities and manoeuvres performed in the sport. Undoubtedly, there is agreement that non-contact ACL injuries frequently occur with opposition players in close proximity (54, 63, 159, 204, 270, 426, 541), with externally focused attention (such as focusing on an opponent, ball, or teammate), under high temporal and visual spatial demands, which increases cognitive loading (291, 310, 318, 426).

Figure 2.4. Example images illustrating inciting event of ACL injury during a COD and jump-landing. Taken from Krosshaug et al. (307) and Stuelcken et al. (541)

2.1.4.1 Kinematic characteristics of inciting events of ACL injuries

Visual and qualitative assessments of non-contact ACL injury have reported observations of large knee valgus motion (ligament dominance) (63, 94, 159, 204, 270, 291, 310, 426, 541, 583), with an extended knee (quadriceps dominance: first degrees of knee flexion) (63, 94, 204, 270, 291, 310, 426, 541, 583), with the foot outside the knee and the body medial to foot, typically with asymmetrical loading (i.e., unilateral deceleration or weight predominantly on one limb: leg dominance)) (63, 204, 270, 291), and lateral trunk flexion towards the stance limb (trunk dominance) (248, 291, 541) (Figure 2.4). These characteristics are all primary factors that can increase knee joint loading, thus ACL strain (39, 126, 127, 183, 242, 273, 301). Internal and external rotation of the tibia has also been observed during ACL injuries (94, 291, 298, 426) which have been shown to increase ACL strain (342-344). Additionally, hip flexion and abduction have been reported during incidences of ACL injury (63, 204, 270, 291) which links to a wide lateral foot plant for plant-and-cut actions. Foot positioning has also been described, with characteristics such as heel strikes (54, 376), foot flat (63), pronation (204, 291), and externally rotated foot positions observed (270). Importantly, studies have reported that the ACL tear typically occurs ≤

50 ms after IC (297, 298, 310), which most likely provides insufficient time for postural adjustment (297). Thus, it has been suggested that neuromuscular "feed-forward" strategies (i.e., optimal lower-limb and trunk initial postures and pre-activation), prior to ground contact, play an important role in non-contact ACL injury mitigation (297).

2.1.4.2 ACL injured vs. control groups

Although the visual and qualitative video-based studies have provided valuable insight into the potential mechanisms of the inciting events of ACL injuries, it is important to note the majority of the abovementioned studies have lacked a CG (94, 204, 270, 291, 310, 426, 541, 583). Therefore, it would be unsuitable to definitively conclude that athletes displaying the abovementioned kinematics associated with ACL injuries will cause an ACL injury, because similar kinematics may have been observed upon inspection of controls.

Review of the literature has revealed four ACL observational studies which have included a CG (54, 248, 376, 509). Hewett et al. (248), using two-dimensional (2D) measures, reported female ACL injured athletes compared to female controls showed less forward trunk lean (p = 0.005, ES = 1.37) and a trend towards higher lateral trunk flexion angles and KAAs versus the control (p = 0.16, p = 0.13 – descriptive data not provided). Similarly, Sheehan et al. (509) found ACL injured athletes displayed less forward trunk lean, with the COM posteriorly further than the base of support. This position has the potential to increase quadriceps activation and anterior tibial shear which can increase ACL strain (42, 43, 343, 344, 615). Montgomery et al. (376) reported in rugby ACL injured athletes demonstrated significantly (p < 0.001) less knee flexion and greater dorsi flexion (p = 0.033) in comparison to the CG. Similarly, Boden et al. (54) revealed significantly lower KFA in ACL injured athletes (versus a CG) when the foot was flat on the ground, thus supporting the quadriceps dominance theory of ACL injury. Boden et al. (54) also revealed significantly greater hip flexion, significantly less plantar flexion, contact with hind foot, and progressive increases in knee abduction motion throughout the contact; demonstrating landings with ankle kinematics of less plantar flexion and more of a rear foot/ flatfoot landing may predispose the athlete to abnormal attenuation of the GRF by increasing the forces experienced at the knee, as the gastrocnemius-soleus complex may be in an inefficient position to attenuate and dissipate the GRFs (452). As such, further research inspecting the kinematics of ACL injury situations with comparisons to a CG is required to improve our understanding of the potential mechanisms of ACL injury.

2.1.4.3 Limitations of visual and qualitative assessment studies of ACL injuries

Generally, ACL injury visual observation studies qualitatively evaluate joint angles and positions, and subjectively describe the motion of the lower-limb and trunk and, in some cases, estimate joint angles and ROM (94, 204, 270, 291, 310, 376, 426, 541, 583). However, Krosshaug et al. (309) has questioned the accuracy of such techniques, reporting substantial errors and underestimations in knee flexion, hip flexion, and knee internal rotation when qualitatively estimating joint kinematics during cutting. It is worth noting that some studies have overcome the limitation of qualitative assessment by measuring the joint angles and motion at IC and during the stance phase (i.e., ImageJ software) for greater accuracy and description of the events (54, 248, 509). Additionally, it should be noted that visual observation studies are generally limited to match footage of elite sports, with little information in recreational level athletes and ACL injuries sustained during training.

Other limitations of visual and qualitative assessments of ACL injuries are the image quality (sample rate and resolution) and viewing angles (307). It is often difficult to capture footage of an athlete in the exact sagittal or frontal plane which increase the susceptibility to parallax or perspective error, while the difficulties in identifying anatomical landmarks in clothed individuals are also problematic (248). Without conducting 3D analysis, it is difficult to describe and assess transverse kinematics of the tibia and femur (616), thus caution is recommended when interpreting and describing coronal movement via visual inspection. Furthermore, description of knee abduction motion does not account for rotation of the leg, thus the KAAs reported could be a combination of the knee abduction, internal rotation of the femur, and external rotation of the tibia (248), while visual inspection is incapable of determining the time course for joint angles, velocities, and accelerations. Furthermore, cautious interpretations regarding the studies that have described the mechanism of injury are recommended, because it cannot be concluded whether the kinematics displayed were causes or consequences of the injury. However, there is strong evidence to suggest that the "high-risk" kinematics displayed by athletes during COD and landing manoeuvres are associated with increased knee joint loading (127, 228, 267, 272, 273, 301, 321, 358, 516, 519), thus risk of injury.

2.1.4.4 Model-based image-matching

In light of the aforementioned limitations of visual and qualitative inspection of joint kinematics, a model-based image-matching technique has been validated to extract joint kinematics from video recordings using one or more uncalibrated cameras (298, 308). Koga et al. (298) applied model-based image-matching to 10 ACL video cases reporting mean flexion angles of 23°, neutral knee valgus at IC but increased 12° 40 ms later, which was accompanied by 8° internal rotation of the knee, then 17° external rotation. Additionally, GRFs of 3.2 times body weight (BW) were estimated, occurring 40 ms after IC. The authors revealed consistent kinematic patterns amongst all cases, with immediate knee valgus motion within 40 ms after IC, tibial internal rotation with limited knee flexion, consistent with the observations of the aforementioned visual and qualitative studies. Consequently, it appears that knee valgus loading is a large contributing factor to ACL injuries, and tibial internal rotation is coupled with valgus motion. Model-based image-matching has been shown to produce greater accuracy compared to visual inspection, therefore further inspection of ACL injury cases using this technique is warranted. However, a note of caution is recommended because the model-based image-matching technique is time-consuming, whereby matching one video sequence has been reported to take 1-2 months (298).

2.1.4.5 ACL injury video inspection summary

Qualitative and quantitative analysis from video footage of ACL injuries has provided a greater insight into the inciting events and kinematic characteristics associated with ACL injuries. There is large agreement that knee valgus motion and loading with limited knee flexion is a contributing factor to ACL injuries, particularly during single-leg weight bearing situations such as CODs and landings. In addition to hip and knee motion, there is evidence to suggest that trunk positioning, ankle, and foot position are connected to ACL injuries and can influence subsequent knee joint loading. Thus, addressing knee and trunk control are likely to be critical strategies for ACL injury mitigation.

Collectively, there is consensus that the majority of ACL injuries occur with opposition players in close proximity, or with externally focused attention (ball, opponent etc), which could alter lower-limb and trunk kinematics and co-ordination, potentially inducing techniques that evoke hazardous knee joint loading (9, 65). Going forward, based on the aforementioned evidence, it could be more suitable to screen athletes' whole-body kinematics during sport-specific situations and scenarios where ACL injuries are likely to occur. For example, unplanned scenarios such as side-stepping past an opponent or catching a ball with the athlete's attention externally focused; though practitioners run the risk of making the task complex and potentially "high-risk" and hazardous (187).

2.1.5 Fatigue failure: repetitive and accumulation of loading as a mechanism of non-contact ACL injury

Although studies have quantified the visual and biomechanical characteristics of the inciting event of non-contact ACL injuries and suggested potential kinematic mechanisms of injury such as knee valgus, extended knee positions, and lateral trunk flexion (52, 63, 94, 170, 291, 298, 376, 426, 583), caution is advised before definitively concluding and establishing the "inciting event" as the sole cause of the injury. Non-contact ACL injuries are complex in nature, influenced by both internal and external factors (48, 365, 469); thus, inspection of the inciting event may only tell a limited part of the story (48).

Injuries occur when the energy transferred to the body exceeds magnitudes or rates which exceeds the tissues threshold (365); however, athletes perform hundreds and thousands of jump-landing and COD manoeuvres without getting injured, so this raises the question: "Why does an athlete all of a sudden sustain an ACL injury?" An underexplored but important factor which may explain the causes of injury are the preceding events prior to the ACL injury (48, 365).

It is commonly assumed that a single catastrophic overload is the mechanism injury (241, 298), attributed to suboptimal biomechanics. Although a single catastrophic overload (particularly multiplanar knee joint loading) which exceeds the ACL's tissue tolerance ruptures the ACL (288, 327, 342, 424, 513), it is important to take into account the ACL loading history of the athlete preceding the injury (324). Herrington and Cooke (234), based on retrospective analysis, found a large proportion of ACL injured athletes subjectively reported a fatiguing event (physical or cognitive) seven days prior to the injury. However, the exact workload preceding the injury was not investigated in contrast to non-injured athletes; thus, further research quantifying the high intensity (COD, acceleration, deceleration) activity profiles of athletes prior to ACL injury is required.

Athletes frequently perform manoeuvres (COD, landings, decelerations) associated with high impact and joint moments without getting injured; however, an accumulation of high magnitudes (potentially due to suboptimal technique) and cyclic and repetitive knee joint loading may lead to "fatigue failure" of the ACL (29, 85, 193, 324, 616). This repetitive and cyclic loading can damage collagen ACL fibres and fibrils (85), reducing the structural integrity of the ACL (85), and can lead to subsequent rupture (29, 85, 193, 324, 616) and thus, an alternative mechanism of non-contact ACL injury. For example, Lipps et al. (324) reported repetitive and high loading (combined 4 × BW with impulsive compression, KFM, and tibial internal moment) resulted in fatigue failure and ACL ruptures in human cadaver specimens. Similarly, Beaulieu et al. (29) found ACL injury risk was eight times greater with limited femoral internal rotation during in vitro repeated pivot landings, while fatigue failure has also been observed in the medial collateral ligament in rabbits (557). Recently, Chen et al. (85) reported repetitive impulsive loading inflicted damage to collagen fibrils and fibres, thus affecting the structural integrity of the ACL. Consequently, it could be argued that performing repetitive loading cycles of manoeuvres such as CODs, landings, and decelerations with the suboptimal technique, thus high knee joint loading (39, 242), may evoke tissue damage which is beyond the ACLs ability to adapt and heal. This process, without adequate rest and repair, could lead to micro damage and subsequent failure (29) from mechanical loads which can no longer be tolerated (20, 85, 161).

It is suggested to reduce the risk of fatigue failure of the ACL by using preventative strategies and focusing on modifiable risk factors including: 1) reducing the magnitude of knee joint loading athletes experience via appropriate injury mitigation training and strength and conditioning (241, 432, 435); and 2) monitoring and periodising the volume and exposures to repetitive and high knee joint loading cycles an athlete experiences in their sport (616); similar to the workload, mitigation, and monitoring approaches adopted in baseball pitchers and cricket bowlers (50). The aims of these strategies are to maintain homeostasis of the ACL, by ensuring the ACL micro damage and degradation (616) is less than or equal to the rate of remodelling, thus reducing the risk of injury. Myrick et al. (399) reported a significant increase in ACL volume (magnetic resonance imaging evaluation) over the course of a season in female soccer players, which may potentially indicate greater strength. However, monitoring ACL volume using magnetic resonance imaging is not feasible for most practitioners and athletes. It is worth noting that published research in this area is preliminary to date, and further research is required exploring fatigue failure and changes in ACL volume as mechanisms of injury (prospective or retrospectively), risk assessment tool, and monitoring changes in response to training intervention strategies (193).

2.2 Injury risk screening

2.2.1 Prospective and matched case-control studies: ACL injury risk factors

It is important to note that the mechanisms of ACL injury and prospective risk factors are different; for example, the mechanisms of injury typically involve descriptions of the actions, behaviours, and biomechanical characteristics of the inciting injury (20, 429), whereas ACL prospective risk factors are variables used for screening and prediction of injury (20, 429). Although prospective risk factors and ACL injury mechanisms are related, they are not identical; thus, the terms should not be used interchangeably (20, 429).

The prospective cohort research design is the strongest method for identifying prospective risk factors for ACL injury (429, 508), while retrospective matched-case control studies, although not as strong as prospective cohort studies, can also provide insight into the potential injury risk factors. Consequently, numerous screening tests and programmes such as the drop vertical jump (DVJ), single-leg squat, single-leg landing, strength assessments, neuromuscular activation, and trunk control, have been used in various athletic populations in an attempt to understand and establish potential ACL risk factors. The following section will primarily focus on the DVJ prospective research studies, while briefly discussing the other assessments and risk factors.

<u>DVJ</u>

Seminal work from Hewett et al. (242) prospectively examined 205 female adolescent (~16 years old) athletes (soccer, basketball, and volleyball) during a DVJ using 3D motion and GRF analysis. Peak

KAM, notably, was found to predict non-contact ACL injury with a specificity of 73% and sensitivity of 78%. Additionally, athletes who sustained an ACL injury displayed 8° greater KAAs, 20% higher vertical GRF (VGRF), 2.5 times greater KAMs, and side-to-side differences in KAMs (6.4 × greater) between limbs compared to uninjured female athletes. As such, it was concluded that knee abduction loading and motion were predictors of injury in female adolescent athletes; however, it should be noted that only nine non-contact ACL injuries were observed, lower than Bahr and Holme's (19) minimum recommendations for detecting moderate to strong associations between a risk factor and injury (20-50 cases). Furthermore, Hewett et al. (242) have also received criticism for a lack of family wise error correction (508) and failing to normalise KAM or VGRF data relative to height or body mass (BM) when comparing between groups (415), whereby research from Norcoss et al. (417) has shown normalisation to influence KAM clinical evaluations. Moreover, the cut-off frequencies (COF) used by Hewett et al. (242) have also been questioned (extreme mismatched of 9 and 50 Hz), as researchers have shown matched and mismatched COFs can affect KAM values and ranking within a cohort (304, 493). However, Roewer et al. (493) advocates the use of mismatched COFs for evaluating KAMs during DVJs.

In contrast to the observations by Hewett et al. (242), three recent studies that used matched COFs have failed to demonstrate KAMs during DVJs can predict non-contact ACL injury in female athletes (311, 321, 525). Krosshaug et al. (311) found no association between DVJ peak KFA, KAA at IC, peak KAM, and peak VGRF with ACL injury in 710 female soccer and handball athletes. However, medial knee displacement was the only variable associated with increased risk of non-contact ACL injury, although there was substantial overlap between the frequency of distribution in medial knee displacement between injured and non-injured athletes and receiver operating characteristics (ROC) revealed an area under the curve (AUC) of 0.6, thus indicating poor-to-failed combined sensitivity and specificity. It is important to note, however, that the sample of Krosshaug et al. (311) were older (~21 years old) than the cohort of Hewett et al. (242) (~16 years old) and were therefore more likely to have performed ACL mitigation training and strength and conditioning, thus potentially explaining some of the differences in results.

Similarly, Leppänen et al. (321) also found DVJ knee valgus angle, KAM, and medial knee displacement failed to predict non-contact ACL injury in 171 female basketball and floorball players. Interestingly, however, stiffer landings (lower KFA) and higher VGRFs were associated with increased risk of injury, but ROC analysis revealed an AUC of 0.6 and 0.7 for peak KFA and VGRF, respectively, indicating a failed-to-fair combined sensitivity and specificity of the test. The results of greater VGRF partially support the findings of Hewett et al. (242), yet the absence of significant associations of frontal plane biomechanics with non-contact ACL injury are in contrast, though Leppänen et al. (321)

attributed this potentially to limited statistical power. Recently, Smeets et al. (525) found no differences between ACL injured and non-injured groups for KFA, KAA, HFA, or KAM during a DVJ task, further questioning the predictive ability of DVJ testing and ACL injury.

In contrast to using 3D analysis, researchers have prospectively examined DVJ landing mechanics using the qualitative assessment LESS (431, 527), though conflicting observations have been demonstrated. Smith et al. (527) found no associations between LESS score and ACL injury ($p = 0.32$) and no predictive value of the LESS in high school athletes ($n = 3876$). Conversely, Padua et al. (431), in a homogenous cohort of youth soccer players, found uninjured athletes displayed lower LESS scores (4.43 ± 1.71 vs. 6.24 ± 1.75, $p < 0.05$, ES = 1.05) compared to ACL injured, and established a cut-off score of 5 which generated a sensitivity (ROC) of 86% and specificity of 64%. However, it should be noted that only seven ACL injuries were observed, four of which were non-contact which is lower than Bahr and Holme's (19) recommendations for screening injury risk factors, and therefore lacks statistical power.

<u>2D knee control</u>

In contrast to previous studies that have screened bilateral landing tasks (242, 311, 321, 431, 527), several studies have screened single-leg biomechanics during slow (472) and high velocity tasks using 2D analysis (133, 418), with a specific focus on dynamic knee control. Dingenen et al. (133) reported summed knee valgus and lateral trunk motion during a single-leg drop landing was greater in athletes who sustained a knee injury (seven injuries, four of which were ACL), compared to injury free athletes, showing an AUC of 0.803 ($p = 0.012$). However, it should be noted that, again, this prospective study was performed in a low sample size ($n = 50$) and recorded a low ACL incidence rate ($n = 4$). In a greater sample size ($n = 291$), female high school basketball players who sustained an ACL injury ($n = 28$) demonstrated greater dynamic knee valgus distance at IC (2.1 ± 2.4 vs. 0.4 ± 2.2 cm, $p = 0.006$, ES = 0.74) and at maximum (8.3 ± 4.3 vs. 5.1 ± 4.1 cm, $p = 0.007$, ES = 0.76) compared to uninjured control athletes (418). It is worth noting, however, that the authors only performed an independent t-test and failed to calculate ROC statistics for sensitivity and specificity. Moreover, basketball and floorball athletes who displayed greater frontal plane projection angles (FPPA) during a single-leg squat were 2.7 times more likely to sustain a lower-extremity injury (472), but there was no significant association between FPPA and knee injury, and ROC indicated poor combined sensitivity predicting lower-extremity or ankle injuries (AUC = 0.57-0.59). These findings were attributed to the substantial overlap in FPPA measures between injured and injured athletes; thus, failing to divide athletes into two distinct groups.

Other potential risk factors

Alternative assessments have also been used to identify ACL injury risk factors, such as time to stabilisation during a backwards hop which was associated with greater risk of non-contact ACL injury (157). Deficits in trunk control and proprioception have also been identified as potential predictors of knee injury status and non-contact ACL injury risk in female athletes but not males (635, 636); however, the ability to directly perform the trunk control tests performed by Zazulak et al. (635, 636) in a field-setting does not appear feasible, and it is worth noting that only six ACL injuries were observed. Additionally, Zebis et al. (637) reported five athletes who sustained an ACL injury demonstrated reduced semitendinosus and increased vastus lateralis activity during the preparatory phases of a cutting task, while Smeets et al. (525) found four ACL injured athletes displayed greater vastus lateralis, greater lateral hamstring and reduced medial hamstring activation around peak ACL loading during a DVJ task. Finally, with respect to the ability of strength measures being able to predict non-contact ACL injury, conflicting observations have been observed regarding the association and predictive capabilities of isokinetic strength measures (287, 394, 528, 536, 567), hip strength measures (287, 536), and leg press strength (536).

2.2.2 Can screening predict injury?

Recently, Bahr (18) has suggested: "No current screening tool can predict injury and probably never will," while more recently, Clarsen and Berge (91) state: "Identifying who will get injured (and who will not) based on risk factor testing is wishful thinking for the foreseeable future." It is important to highlight that the mechanisms of injury are multifactorial, involving a combination of both intrinsic and extrinsic factors (48, 365). As such, due to the complexity of the mechanisms of ACL injury (469), it would be inappropriate to definitively conclude that a certain risk factor can predict injury (365). Furthermore, in contrast to the inciting event of injury (action), it is thought that the events preceding the injury, such as amalgamation of several factors (i.e., rapid spike in workload, neuromuscular fatigue, reduced conditioning) causes a chain of shifting circumstance that contributes to the injury (48, 161, 365).

Although previous studies have reported a strong association between a risk factor and ACL injury (i.e., p value or odds ratio), it is important to note that association does not equate to predictive capacity, and typically these risk factors do not have the capacity to predict injury with sufficient accuracy (18). Thus, predictive tests require a higher standard of statistical testing such as ROC/ AUC. Moreover, studies that have shown promising results regarding the predictive capabilities of a specific risk factor (157, 242, 287, 431, 635, 636) are specific to that population only, and therefore may have limited generalisation and application to athletes from different athletic populations, sexes, and ages.

2.2.3 Limitations of prospective and matched case-control studies

It is worth acknowledging that the biomechanical or strength characteristic used for prospective screening is only representative of the time of testing and is temporal, and therefore likely to change throughout the season due to changes in maturation, strength and conditioning, neuromuscular training, and match demands (311, 365, 536, 575). However, changes in these characteristics throughout the follow-up period of prospective investigations are rarely monitored. Prospective studies have reported inconsistences in the follow up periods and the mean time of injury, which varies substantially within and between studies (Table 2.1). Researchers have reported ACL injures occur within days following screening to as long as several years (Table 2.1). It would therefore seem unreasonable to assume the mechanics or strength characteristics displayed by an athlete during initial screening (e.g., baseline or pre-season) are likely to be similar several years later at the time of injury, and highly unlikely that the risk factor (score) will be similar too (311, 536, 569, 575).

Table 2.1. Follow-up periods and time of non-contact injuries in prospective screening research

Study	Follow-up period	Time of injury (since baseline screening)
Hewett et al. (242)	Screened for 2 autumn and 1 winter season	Mean: 5.0 months (range: 0.6-13.1 months)
Smith et al. (527)	3 years	Mean: 224 ± 150 days (Range: 1-434 days)
Steffen et al. (536)	Up to 8 years	Mean: 1.8 ± 1.8 years
DuPrey et al. (157)	1 to 4 years	Mean 19.0 ± 10.7 months (Range: 6.8-44.1 months)

Note: Several studies provide follow up duration (typically 1-3 years) but do not specify time of injury (Leppänen et al. (321), Zazulak et al. (635, 636), Khayambashi et al. (287), Dingenen et al. (133), Numata et al. (418)

Another issue in prospective research is that the reliability of the risk factor is rarely established; thus, it unclear that an athlete’s clinical evaluation (magnitude and subject ranking of variable) would be similar between sessions for a specific variable. As stated previously, a large proportion of prospective studies have had low sample sizes, and the incidents of ACL injury, again, are generally lower than Bahr and Holme’s (19) recommendations for detecting moderate to strong associations between a risk factor and injury (20-50 cases), thus lacking statistical power. Finally, a key issue in predicting injury risk is the overlap between injured and uninjured athletes for the variable (test score) in question, with tests failing to divide injured and uninjured athletes into two clear distinct groupings (Figure 2.5). This overlap is problematic, which therefore makes it difficult to establish an appropriate cut-off score or value for defining “high-risk” athletes (18, 372, 472).

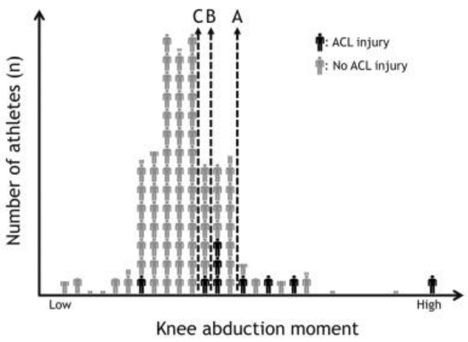


Figure 2.5. Schematic representation of Hewett et al. (242) data illustrating KAM and risk of ACL injury. Taken from Bahr (18)

2.2.4 Advantages and rationale for screening

Although it is difficult to ascertain whether movement screening can accurately predict which athlete will become injured (18, 311, 372), it does not mean that movement screening is redundant and obsolete (91, 243, 354, 372, 569). In fact, movement screening serves a wide range of purposes and its usefulness should not be dependent on its ability to predict injury. For example, some of the positive benefits of movement screening are as follows (243, 354, 372, 436):

1. Identify specific biomechanical and neuromuscular deficits to subsequently inform and provide individualised and targeted interventions
2. Medical/research team engagement with athletes
3. Potential placebo and Hawthorne effects
4. Athlete education and team awareness
5. Objective testing that can monitor and determine the effectiveness of training interventions
6. Enhancement of athlete performance (injury mitigation programmes are mostly unharmful and, in some cases, can improve athletic performance such as jumps, hops, sprinting – depending on athletes' training status).

It is important to note that the results of movement screening may not accurately predict who will get injured (18, 311, 372); however, biomechanical and neuromuscular deficits that are identified, can provide an indication of the relative risk of injury (569) and an athlete "injury risk profile" can be created (240). As such, it is important that movement screen tests are still included in the sports medicine programmes of sports teams and institutions, while the use of movement screening to determine the effectiveness of a training intervention on potential risk factors associated with injury (i.e., effect of training on drop-landing KAMs) play a critical role for research purposes (372).

2.2.5 Field-based screening tools

3D motion analysis is considered the gold standard for screening and assessing athletes' movement kinetics and kinematics (187, 241); however, this method is complex, time-consuming (testing one subject can take ~2 hours), and expensive, which is often restricted to laboratory settings, and requires trained personnel to conduct the testing (163, 241, 302, 408, 527); therefore, having limited applicability to practitioners working with athletes in "real-word" environments. Due to these limitations, testing large cohorts and performing large-scale mass screening is difficult for practitioners.

In light of the limitations associated with 3D motion analysis, an emerging area in injury risk screening is the development of field- and clinical-based screening tools. Over the last decade, a number of screening tools and methods have been created in an attempt to identify movement deficiencies associated with increased risk of injury, such as the LESS (434), TJA (397), QASLS (237), 2D analysis (i.e., FPPA) (607), and observational screening (408) (Table 2.2). Athletes that display suboptimal mechanics, using the more time-efficient and user friendly screening tools, can then be targeted with individualised training interventions to help reduce the relative risk of injury (187, 372).

In order for a screening tool to be successful for identifying potentially "high-risk" athletes in field-settings, Fox et al. (187) identified several important considerations that practitioners and clinicians should acknowledge before investing their time and financial resources into screening. The considerations were divided into appropriateness, technical adequacy, predictive validity and usability, with specific considerations identified as follows:

1. Prospective evaluation/ predicting injury
2. Validity
3. Intra- and inter-rater reliability
4. Cut-off values established for individuals identified as "at-risk"
5. Equipment
6. Rater and athlete training considerations to carry out and perform testing
7. Duration of testing and analysis
8. Screening method applicable across multiple sports and athletes

While 2D assessments of FPPA demonstrate strong relationships with 3D measures of knee valgus during single-leg squats (r = 0.78-0.79) (211, 233) and single-leg landing (r^2 = 0.72) (529), and good reliability (97, 99, 211, 236, 386, 387, 529, 607, 608), due to parallax error, however, the data is unlikely to be homoscedastic, thus inappropriate statistical approaches are often used. Furthermore, 2D methods still require additional software and time to analyse data and is also susceptible to perspective error. Knee observational screening has also been performed (163, 408, 537), but given the mechanistic connection between the trunk, hip, knee, and ankle with "high-risk" movement patterns and knee joint loading (138, 141, 241, 245, 391), the primary focus of this section is to discuss current qualitative screening tools which evaluate whole-body movement patterns, and to discuss the advantages and disadvantages of current qualitative field-based screening tools (LESS, TJA, and QASLS) available to practitioners and clinicians (Table 2.2).

Landing error scoring system (LESS)

The traditional LESS is a DVJ assessment whereby lower-limb and trunk mechanics are qualitatively assessed against a grading criteria using video cameras placed in the sagittal and frontal plane (Table 2.2) (434). Padua et al. (434) validated the LESS to 3D motion analysis reporting athletes that scored higher LESS scores (i.e., poor jump-landing technique) exhibited kinetic and kinematics associated with greater risk of ACL injury, such as decreased hip and knee flexion (sagittal plane kinematics), increased knee valgus and hip adduction angles, and greater frontal and transverse hip and knee moments. However, conflicting evidence has been observed regarding the ability of the LESS to predict ACL injury (Table 2.2) (431, 527), and no evidence to support its ability to predict lower-extremity injury (265). From a reliability perspective, moderate to excellent intra-and inter-rater reliability measures have been reported for the LESS (431, 434, 527) (Table 2.2), and no significant differences ($p > 0.05$, r = 0.587-0.611, $p < 0.001$) in LESS scores between different landing surfaces have been reported (263). While more recently, the LESS has been modified by using real-time immediate scoring (430) and the iLESS has been developed whereby athletes are classified as "high-" or "low-risk" based on one trial viewed in the frontal plane (107).

The relative simplicity of the LESS for athletes to perform, and subsequent rating, has led to large-scale mass screening in various populations (soccer, high school athletes, military) (41, 431, 527), and the minimal equipment requirements are further strengths of the assessment. However, the LESS does have its drawbacks: firstly, the standardised 30 cm box height is a limitation because it may not be reflective of the landing heights for some athletes and thus may be too easy for athletes who can jump higher than 30 cm (476); secondly, the bilateral nature of the task and the fact that a "bounce" is examined, has been questioned regarding its ecological validity to the scenarios and actions of ACL injuries (302, 372) typically, unilateral landings and COD manoeuvres; and thirdly, it is worth noting that the "bounce" during the first landing of the DVJ is biomechanically different to the second landing of the DVJ (25).

Table 2.2. Summary of current qualitative screening tools currently used by practitioners and researchers

	LESS	TJA	QASLS
Description	Padua et al. (434) developed an assessment whereby lower-limb and trunk mechanics are assessed during the first landing of a DVJ, using video cameras placed in the sagittal and frontal plane. Athletes jump 50% of their height away from a 30 cm box and rebound for maximum height, and are then scored against a 17 point criteria at initial ground contact (trunk, hip, knee, and ankle flexion angles at IC; knee valgus angle at IC; lateral trunk flexion angle at IC); foot position at IC and the time between IC and maximum knee flexion (stance width—wide or narrow; foot position—toe in or out; symmetry of initial foot contact); lower-extremity and trunk movements between IC and maximum knee flexion (knee flexion and valgus displacement; trunk and hip flexion at maximum knee flexion); overall sagittal plane movement; and the general perception of landing quality, which are suggested to be indicative of faulty biomechanics associated with increased ACL injury risk (434). The LESS has been modified by using real-time immediate scoring (430) and the iLESS has been developed whereby athletes are classified as "high-" or "low-risk" based on one trial viewed in the frontal plane (107).	Myer et al. (397) developed a repeated plyometric jump task assessment, whereby athletes perform repeated tucking jumps (knees to chest) over 10 seconds and involves the subsequent grading of ten 'flaws' and dichotomous items. The ten scoring items are split into categories, with three criteria relating to knee and thigh motion (lower-extremity valgus at landing; thighs reaching parallel; unequal thighs side-to-side), four criteria relating to foot position during landing (foot placement shoulder width apart; foot placement not parallel; unequal foot contact timing; excessive landing contact noise), and three criteria relating to plyometric technique (pauses between jumps; technique declines prior to 10 seconds; does not land in same footprint). Two cameras can be used to retrospectively grade technique; however, the assessment can also be performed in real-time.	Herrington et al. (237) developed the qualitative analysis of single-leg loading used for screening lower-limb and trunk characteristics and motion during slow velocity single-leg squats and single-leg drop landing tasks (8, 120, 235). The scoring system, adapted from previous studies (111, 600), involves a dichotomous scoring of movement quality which has been subdivided into six categories: arm strategy, trunk alignment, pelvic plane, thigh motion, knee position, and steady stance (8, 237). Athletes are filmed in the frontal plane and retrospectively screened against the scoring criteria out of 10, with higher scores being indicative of poor technique and potentially increased risk of injury (8, 120, 235, 237); however, scoring can also be performed real-time (120).
Reliability	**Intra-rater** • Padua et al. (434): Total score ICC = 0.91, SEM = 0.42 • Onate et al. (427): item reliability k = 0.459-0.875 • Smith et al. (527): Total score ICC = 0.92-0.97 **Inter-rater** • Padua et al. (434): ICC = 0.84, SEM = 0.72-0.79	**Intra-rater** • Myer et al. (397): r = 0.84 • Herrington et al. (238): k = 0.86-1.00, PEA = 87.2-100% • Read et al. (482): Total score ICC = 0.88 • Fort-Vanmeerhaeghe et al. (182): Average PEA = 90.8%, Total score ICC = 0.94-0.96 • Dudley et al. (155): Total score ICC = 0.42-0.72	**Intra-rater** Almangoush et al. (8): k = 0.89-1.00, PEA = 95-100%, **Inter-rater** Almangoush et al. (8): k = 0.63-1.00, PEA = 83-100% **Between-session** Dawson et al. (120): Total score ICC = 0.91, SEM = 0.89, SDD = 2.49

	• Padua et al. (430): Real-time ICC = 0.81, SEM = 0.72-0.79 • Onate et al. (427): Total score ICC = 0.835 • Cortes et al. (107): iLESS = k = 0.600-0.692, PEA = 80-90% Jacobs et al. (263): No significant differences (p > 0.05, r = 0.587-0.611, p < 0.001) in LESS scores between different landing surfaces.	**Inter-rater** • Herrington et al. (238) k = 0.88, average PEA = 93% • Fort-Vanmeerhaeghe et al. (182): Average PEA = 92.1%, Total score ICC = 0.94, 9 of 10 items $k \geq$ 0.65 • Dudley et al. (155): p < 0.001 in scores between raters, ICC between raters ICC = 0.47 • Mayhew et al. (350): k = 0.46-0.86 across variables, total score weighted k = 0.62-0.80 • Gokeler & Dingenen (196): Total score ICC = 0.85-0.88, SEM = 0.8-1.0, SDD = 2.3-2.9; k = 0.09-0.88 for items **Between-session** • Read et al. (482): ICC = 0.53-0.55, TE = 0.90-1.01, knee valgus only variable to achieve substantial agreement k = 0.67-0.78 • Gokeler & Dingenen (196): Total scores ICC = 0.93-0.96, SEM = 0.5-0.7, SDD = 1.3-2.1; k = 0.19-0.88 for items – modified TJA	
Validity	Padua et al. (434) validated the LESS to 3D motion analysis reporting athletes that scored higher LESS scores (i.e., poor jump-landing technique) exhibited kinetics and kinematics associated with greater risk of ACL injury, such as decreased hip and knee flexion (sagittal plane kinematics), increased knee valgus and hip adduction angles, and greater frontal and transverse hip and knee moments. Onate et al. (427) showed strong associations between qualitative assessments and 3D kinematics.	Good face validity (391); the specific grading criteria have been divided into 5 criteria which are suggested to be representative the biomechanical and neuromuscular imbalances of ACL injury including ligament dominance, quadriceps dominance, leg dominance (residual injury deficits), trunk dominance (core dysfunction), and technique perfection, with higher scores representing poorer jump-landing technique and thus greater risk of injury, but no empirical evidence to support it. No evidence to confirm its ability to identify athletes who exhibit high knee joint loads.	The QASLS has been validated against 3D motion analysis observations of single-leg motion, with excellent agreements (k = 0.90-0.97, PEA = 95.6-100%) in QASLS scores, and in particular trunk positioning and knee valgus motion single-leg squatting and landing tasks (235). No evidence to confirm its ability to identify athletes who exhibit high knee joint loads.
Prospective research	• Smith et al. (527): no predictive capabilities (p = 0.32) regarding ACL injuries in male and female high school athletes.	• Only FPPA during a TJA has been shown to be associated with increased risk of lower-extremity injury in U11-U12 male soccer players (482).	• Injured athletes (n = 7) displayed lower QASLS scores compared to uninjured (n = 23) (p < 0.001) (256). • No evidence, yet, to confirm its ability to predict non-contact ACL injury.

	• Padua et al. (431): LESS scores above 5 predicted ACL injury in youth soccer players specificity of 86% and sensitivity of 64%. • James et al. (265): No differences in LESS scores (p = 0.83) between injured and uninjured athletes, based on lower-extremity injuries.	• No evidence, yet, to confirm its ability to predict non-contact ACL injury.	
Advantages	• Simple, cost-effective • Permits large mass screening • Applicable to jump-landing sports • Minimal equipment – use of two cameras or real-time • Quick assessment	• Simple, cost-effective • Permits large mass screening • Repeated efforts more challenging than traditional LESS • Applicable to jump-landing sports • Minimal equipment – use of one or two cameras or real-time assessments • Quick assessment	• Simple, cost-effective • Permits large mass screening • Unilateral assessment: potentially increased specificity to unilateral mechanism of ACL injury • Can identify differences between limbs • Applicable to jump-landing sports • Can be performed in sagittal, frontal and transverse plane • Minimal equipment – use of one or two cameras or real-time assessments • Quick assessment
Disadvantages	• Standardised box height may not reflect max jumping capabilities of athletes. • Maybe too easy for athletes who can jump higher than 30 cm. • Bilateral nature may lack ecological validity to the scenarios and actions of ACL injuries. • "Bounce" action results in different biomechanical outcomes measures compared to landing mechanics (25), and may lack ecological validity to landing mechanism of injury.	• Bilateral nature may lack ecological validity to the scenarios and actions of ACL injuries. • Jumping rates can differ, and athletes may adopt pacing strategies. • Test may bias/penalise athletes who display higher jumping rates because of more opportunities to score a flaw. • "Bounce" action results in different biomechanical outcomes compared to landing mechanics (25), and may lack ecological validity to landing mechanism of injury.	• Unilateral tasks typically performed in sagittal plane.

Key: DVJ: Drop vertical jump; ICC: Intraclass correlation coefficient; SEM: Standard error or measurement; SDD: Smallest detectable difference, PEA: Percent exact agreement; ACL: Anterior cruciate ligament; FPPA: Frontal plane projection angle; LESS: Landing error scoring system; TJA: Tuck jump assessment; QASLS; Qualitative analysis of single leg loading; 3D: Three-dimensional; TE: Typical error; max: Maximum; IC: Initial contact

Tuck jump assessment (TJA)

The TJA, developed by Myer et al. (397), is considered "clinician friendly" whereby athletes perform repeated tuck jumps over 10 seconds and involves the subsequent grading of ten "flaws" and dichotomous items (linked to biomechanical and neuromuscular deficits) (391), with higher scores representative of poorer technique (Table 2.2). The TJA, however, has recently been modified to grade the severity of dysfunction (changed from 0 to 1, to 0 to 2 – none, small, and large) (182). The task is thought to be more challenging than typical drop-landing tasks from boxes, with increased repetitions and the addition of fatigue an additional strength of the test (238, 391, 476). Furthermore, the TJA overcomes the issue of standardising the box drop height associated with DVJ (LESS) testing, and reflects the maximal jumping and subsequent landing mechanics an athlete is likely to encounter in sport (476), though this is still a "bounce" action. Specifically, a cut-off score of ≥ 6 has been suggested by Myer et al. (397) for specific targeted interventions for athletes; however, there is no empirical evidence to substantiate this recommendation (350). Furthermore, the TJA has yet to be validated against 3D motion analysis, and there is no evidence to support its ability to predict ACL injury or lower-extremity injury. Only FPPA during a TJA has been shown to be associated with increased risk of lower-extremity injury in U11-U12 male soccer players (482).

In terms of intra- and inter-rater reliability, the TJA, in general, is viewed as moderate to excellent (Table 2.2) (182, 196, 238, 350, 397, 476), though Dudley et al. (155) reported poor inter-rater reliability measures for the TJA scores, but these improved for session 2 indicating a potential learning effect. This discrepancy, however, could be attributed to the limited experience the raters had with the screening tool in contrast to more experienced raters of other research investigations (182, 238, 397, 476). Uniquely, Read et al. (476) examined the between-session reliability of the TJA in male soccer players and found no meaningful differences in TJA scores between sessions (change in mean ≤ 0.24); however, the identified flaws were inconsistent between sessions, with knee valgus the only criteria to achieve good agreements between sessions (Table 2.2). Similarly, Gokeler and Dingenen (196) also reported excellent between-session and good inter-rater reliability for total score during the modified TJA in recreational athletes, but considerable differences in between-session and inter-rater reliability were demonstrated for the individual items (Table 2.2). As such, these findings indicate that soccer players and recreational athletes display similar TJA scores between sessions, but the way the athlete performs the movement, and subsequent deficits, differ between sessions. Furthermore, Read et al. (476) reported a typical error (TE) of 0.90-1.01 for the TJA total score, indicating changes > 1 were needed to be interpreted as "real", whereas Gokeler and Dingenen (196) reported a smallest detectable of difference (SDD) of 1.3-2.1 and 2.3-2.9 based on between-session and inter-rater reliability, respectively.

It is worth noting the TJA has several limitations: firstly, the number of jumps an athlete performed during the 10 second period can vary between athletes (476), athletic populations (526), and sessions (476), thus the jumping rates may positively or negatively affect the overall score. For example, an athlete who performs higher jump frequencies may be at a disadvantage due to the increased number of jumps that can be graded and potentially display a deficit (typically a deficit displayed two or more times results in score awarded), whereas an athlete may display a pacing strategy to subsequently result in lower jump frequencies and subsequent evaluative opportunities to be penalised. Secondly, the bilateral nature of the task, similar to bilateral drop landing, may lack specificity to unilateral actions of non-contact ACL injury (302, 372), and also a "bounce" style of landing mechanics is examined.

<u>Single leg loading (QASLS)</u>

Overcoming the limitations associated with bilateral screening tasks, unilateral single-leg loading tasks such as single-leg squats and single-leg drop-landings can be evaluated due to the greater specificity to non-contact mechanisms of ACL injury and also to examine differences between limbs. Herrington et al. (237) has recently developed the qualitative analysis of single-leg loading screening tool (QASLS) which is typically used for screening lower-limb and trunk characteristics and motion during slow velocity single-leg squats and higher velocity single-leg drop-landing tasks (8, 120, 235). Athletes are typically filmed in the frontal plane and retrospectively screened against the scoring criteria out of 10, with higher scores being indicative of poor technique and potentially increased risk of injury (8, 120, 235, 237); however, scoring can also be performed real-time (120) (Table 2.2).

The QASLS has been validated against 3D motion analysis observations of single-leg motion, with excellent agreements (PEA = 95.6-100%, k = 0.90-0.97) in QASLS scores, and in particular trunk positioning and knee valgus (Table 2.2) (235). In addition, very good to excellent intra- and inter-rater reliability measures have been observed in QASLS scores during single-leg squatting (8), while excellent between-session reliability (Table 2.2) has also been observed with changes in QASLS scores greater than ~2.5 considered meaningful (120). Furthermore, promising findings from an injury prediction perspective have been reported, with individuals that sustained an injury (n = 7) displaying significantly poorer scores compared to uninjured individuals (n = 23) (p < 0.001) (256); however, it is worth noting that this was a low sample size and the specific injuries were not described. As such, further research is required to determine the efficacy of the QASLS in predicting lower-limb injury in greater sample sizes, in particular ACL.

2.2.6 Limitations of qualitative assessments

An issue with qualitative screening tools is the subjective nature regarding the assessment of motions, positions, and kinematics displayed during the movement (187, 353). Krosshaug et al. (309) found experienced observers in ACL injury and visual inspection research reported inaccurate estimates of joint kinematics during a cutting task, typically underestimating knee and hip flexion, and hip and knee internal rotation angles. Furthermore, the experience and competency of the raters grading the movement quality can also influence outcomes and reliability (155, 182, 187, 238). Thus, it is imperative that clear guidelines are established for optimal and suboptimal movement quality which has been consensually agreed by all coaches, practitioners, and potential raters within a team or institution. Furthermore, it is essential that all raters receive adequate training when using screening tools to ensure accurate, consistent, and reliable evaluations are made (187, 353). Finally, coaches and practitioners must look beyond the score and examine the specific deficits and flaws that an athlete demonstrated so informed and targeted interventions can be created and implemented.

2.2.7 Screening cutting movement quality

Although assessments such as the LESS, TJA, and QASLS are simple and easy to use, these assessments, however, generally assess landing mechanics during vertically orientated tasks, which may lack specificity to the multiplanar cutting mechanism of non-contact ACL injury (185, 187, 278, 372). Additionally, it is questionable whether these generic jump-landing tasks (i.e., bilateral drop landing LESS) can identify athletes displaying "high-risk" and "faulty" mechanics during sport-specific tasks (185, 187). In particular, it is inconclusive as to whether assessing landing mechanics can identify athletes who display poor cutting mechanics (86, 279, 302, 421, 551). This finding is concerning for practitioners that want to identify potentially "high-risk" athletes where cutting actions are frequently performed and the primary action associated with non-contact ACL injury in sports such as rugby (376), handball (426), soccer (583), and American football (270). As movement biomechanics and subsequent "injury risk profiles" are task dependent (109, 160, 185, 224, 279, 302, 349, 421, 551), it would be more suitable to assess movement mechanics associated with the primary action of injury in that sport (185, 187, 372).

Despite cutting manoeuvres being cited as a primary action associated with non-contact ACL injury in numerous sports (94, 270, 291, 376, 426, 583), it is surprising that there are lack of screening tools available for practitioners to assess cutting movement quality, in contrast to the several jump-landing screening tools available (i.e., LESS, TJA, QASLS). McLean et al. (361) initially evaluated 2D estimates of frontal plane knee motion during cutting against the gold standard of 3D and found 2D estimates correlated well with side-step ($r^2 = 0.58$) and side-jump ($r^2 = 0.64$) 3D valgus angles, but poorer associations were observed with 180˚ turn knee valgus angle ($r^2 = 0.04$), thus highlighting the difficulty in assessing 2D valgus motion in the frontal plane using a single camera during vigorous CODs. Despite the promising relationship observed between 2D and 3D knee valgus angles by McLean et al. (361), this method for evaluating cutting mechanics has not been widely adopted. This discrepancy could be attributed to the method only assessing knee valgus motion, which fails to evaluate a range of other technical factors (trunk, hip, foot positioning) associated with increased knee joint loading during cutting (127, 183, 272, 301, 358). Hamdan et al. (218) has recently reported reliable 2D assessments of hip and KFAs during side-step cutting, but did not assess frontal plane or trunk kinematics.

In light of the limitations of McLean et al. (361) and Hamdan et al. (218), Weir et al. (595) demonstrated 2D measures of knee flexion, trunk flexion ROM, and mid-pelvis to foot displacement during unanticipated 45° side-steps were good predictors of KAMs, explaining 55.7% of variability. Expanding on their preliminary work, Weir et al. (593) has recently demonstrated that 2D measures of dynamic knee valgus angle, KFA at foot-strike and ROM, and trunk flexion ROM, when inserted into regression equations, can be used to predict 3D peak KFMs, KAMs, and KIRMs ($p \geq 0.331$, $d \leq 0.34$) during unanticipated side-steps. Although the 2D methods have demonstrated promising findings, the 2D method requires additional time (i.e., manual digitisation) and software to measure knee valgus and other characteristics, thus potentially limiting its applicability in field-settings.

In light of the issues associated with 2D analysis, Jones et al. (278) have recently developed the Cutting Movement Assessment Score (CMAS), which is a qualitative screening tool that assesses cutting movement quality and specific lower-limb and trunk characteristics that are associated with peak KAMs (127, 189, 228, 267, 272, 273, 301, 358, 516, 519, 593), such as PFC braking strategy, and trunk, hip, knee, and foot positioning and motions. In the preliminary study, a strong relationship between CMAS and peak KAM ($\rho = 0.633$; $p < 0.001$) was demonstrated, while moderate to excellent intra-and inter-rater agreements for all CMAS variables (intra-rater $k = 0.60$-1.00, 75-100% agreements; inter-rater: $k = 0.71$-1.00, 87.5-100% agreements) was observed, although lower inter-rater agreements for trunk positioning were observed ($k = 0.40$, 62.5% agreement). Thus, the CMAS may have the potential to identify athletes displaying "high-risk" cutting mechanics. It should be noted, however, that the study was preliminary, containing a small sample size ($n = 8$ subjects, 36 trials) and must be expanded in a greater sample size to confirm its validity. Furthermore, the authors recommended an additional camera to be placed at 45° relative to the COD and using a higher video capture rate (≥ 100 Hz) to permit more accurate and reliable assessments for frontal and transverse plane technique deficits (i.e., trunk positioning, knee valgus).

In contrast to 2D and qualitative analysis, attempts have been made to validate a single Microsoft Kinect V2 depth-sensing camera (30 Hz) against 3D motion (100 Hz) analysis for side-step cutting. Eltoukhy et al. (164) reported the Microsoft Kinect V2 resulted in slight underestimations in cutting kinematics but generally good agreements, consistency, and correlations for sagittal plane cutting kinematics between methods were observed with acceptable limits of agreement (LOA). Conversely, lower reliability, agreements, consistency, correlations, and unacceptable LOA were observed for frontal plane hip and knee kinematics. This finding is concerning because frontal hip and knee kinematics are critical factors related to greater knee joint loads (301, 357); thus, questioning the Kinect V2s ability to identify athletes who display aberrant frontal plane knee and hip motion. Additionally, only ROMs were examined, with IC and peak values left unexamined, while agreements and temporal differences between devices across the whole waveforms were not examined. Further research with greater sample sizes and further technological development (i.e., higher resolution and advanced tracking) is needed to refine the performance of the Kinect V2.

2.3 Change of direction biomechanics

2.3.1 Biomechanics and types of directional changes

A COD is defined as "a reorientation and change in the path of travel of the whole-body COM towards a new intended direction" (118, 621), and the ability to change direction rapidly is an important action associated with successful performance in multidirectional sports (51, 171, 283, 409, 487, 510, 549, 628).

A COD action can be sub-divided into four phases (11, 56, 147, 148, 229, 410):

1. Initial acceleration (positive acceleration)
2. Preliminary deceleration (negative acceleration: to reduce momentum into the COD over PFC and prior steps)
3. Cut/COD (manipulation of base of support relative to COM, to redirect and propel COM)
4. Reacceleration

Changing direction is a multi-step action with the steps preceding and following push-off involved in facilitating effective directional changes (147, 148, 273). Crucially, COD requires manipulation of the base of support relative to COM to create an external braking and propulsive force into the intended direction of travel (Figure 2.6); this typically occurs during the main plant foot contact (e.g., lateral foot plant) (90). It is worth acknowledging, however, that the preliminary deceleration and redirection requirements during directional changes will be governed by the approach velocity, intended COD angle, sporting scenario (i.e., pre-planned, offensive, or defensive agility), and the athletes' physical capacity (neuromuscular control and ability to rapidly produce force) (147, 148).

The "cutting action", defined as a directional change from a few degrees to 90˚ (11), is commonly performed in sports (51, 184, 283, 376, 487, 549, 632), and the execution of such actions can vary substantially between individuals and contexts of sports (11). Three primary cutting techniques have been studied within the literature (39, 60, 102, 289, 407, 462, 470, 473, 547, 548, 562, 563, 568): the side-step, XOC, and split-step, which are typically performed in multidirectional sport and training.

- Side-step cuts are described as an athlete planting their foot laterally opposite to the direction of travel (Figure 2.6a) to create a propulsive impulse into the new intended direction. The body is typically rotated towards the intended direction of travel, and the athlete accelerates towards the direction opposite of the planted leg (11, 94, 473, 548).
- The crossover cut (XOC) (Figure 2.6b) involves positioning the plant foot on the same side (ipsilateral) of the new direction (or sometimes medially across the pelvic midline) and then crossing the opposite leg (contralateral) in front of the body for the new step in the new direction, accelerating in the same direction of the push-off leg (11, 94, 473, 548, 603).

- The split-step (Figure 2.6c) involves the athlete performing a small jump (amplitude jump) prior to push-off, landing with both feet greater than or equal to shoulder width apart, and then, upon landing, the contralateral limb is used for push-off into the intended direction of travel (60, 102, 184).

It is worth acknowledging, however, the three cutting techniques display biomechanical differences which have their own distinct advantages and disadvantages from both performance and risk of injury perspectives (Table 2.3) (39, 60, 102, 289, 407, 462, 470, 473, 547, 548, 562, 563, 568). A full review outlining the biomechanical differences between cutting techniques has been published (146)(Appendix 8.1), thus a brief overview is presented in Table 2.3.

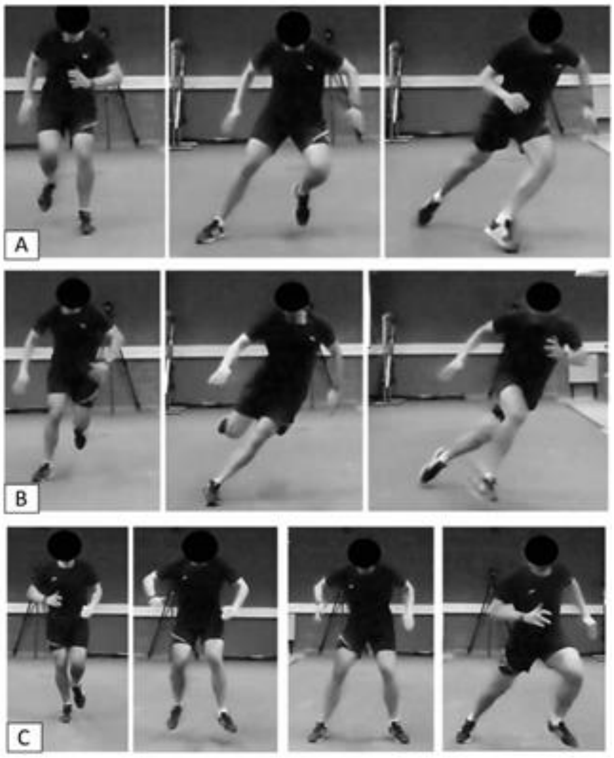


Figure 2.6. Photo sequences of the three cutting techniques: A) Side-step; B) Crossover cut; C) Split-step

<u>Turning</u>

For the purpose of this thesis, turning is synonymous with pivoting and refers to sharper directional changes, typically directional changes ≥ 135˚ such as the modified and traditional 505. A pivot or turn is typically a bilateral turning strategy where one foot rotates and remains in contact with the ground (typically for directional changes ≥ 135˚).

Table 2.3. Advantages, disadvantages, and practical applications of the side-step, crossover cut, and split-step cutting techniques

Cutting technique	Evidence	Advantages	Disadvantages	Practical applications to sport, COD speed, or agility training
Side-step	Besier et al. (39); Suzuki et al. (547, 548); Queen et al. (470); Rand & Ohtsuki (473); Kim et al. (289); Potter et al. (462); Cochrane et al. (94); Montgomery et al. (376); Andrews et al. (11); McGovern et al. (356); Kristianslund et al. (300); Fong et al. (178)	• Sharper angled cut executed vs XOC • Key action for ML propulsion • Successful manoeuvre in deceiving and feinting opponent(s) via lateral foot plant (false step) and trunk and head positioning • Faster than split-step for pre-planned COD speed tasks • Faster approach velocities and shorter preparation times vs. split-step during evasive actions	• Reduced exit velocity vs. XOC • Longer GCT vs. XOC • Propensity to generate large KAMs and KIRMs which increases ACL strain (performance-injury conflict) • Greater incidents of non-contact ACL injuries vs. XOC • Greater medial foot loading (foot pressure) vs. XOC • Risk of lateral ankle sprain injury due to foot and ankle kinematic postures (internal rotation, inversion, supination)	• Attacking agility – 1 vs. 1 situations in sport to get past an opponent or to get into space • Evasive manoeuvres in sport to feint and deceive an opponent e.g., tackle break success in rugby, American football, soccer, etc. • Situations when sharp cut and ML propulsion is warranted
XOC	Besier et al. (39); Suzuki et al. (547, 548); Queen et al. (470); Rand & Ohtsuki (473); Kim et al. (289); Potter et al. (462); Andrews et al. (11); McGovern et al. (356); Wade et al. (579)	• Greater maintenance of velocity during COD vs. other COD actions • Greater exit velocity vs. side-step • Shorter GCT vs. side-step • Potentially reduced risk of sustaining non-contact ACL injury due to the reduced KAM loading	• Acute angled cut executed vs. side-step • Limited deception or feinting – potentially limited application from evasive perspective • Propensity to generate knee varus moment and load lateral component of knee • Greater lateral foot loading (foot pressure) vs. side-step • Risk of medial ankle sprain injury due to foot and ankle kinematic postures (external rotation, eversion, pronation)	• Situations where velocity maintenance and momentum with a subtle COD is warranted, such as collision sports i.e., to break through tackles in rugby and American football • Pre-planned COD tasks in sports such as running around the bases in softball and baseball • Pre-planned COD speed tests where completion time is fundamental; especially when acute cuts are performed
Split-step	Trewartha et al. (562, 563); Nieminen et al. (407); Uzu et al. (568); Connor et al. (102); Bradshaw et al. (60); Munro et al. (389)	• Amplitude jump prior to push-off engages SSC in both limbs • Difficult for opponent(s) to anticipate kinematic cues early • COD is executed late during manoeuvre– choice of two directions • Bilateral strategy dissipates forces and loading across two limbs • Lower knee transverse and abduction loading vs. to side-step • Potentially greater lateral velocity vs. side-step	• Longer preparation time in order to execute COD vs. to side-step and XOC • Most likely reduced approach velocity prior to COD • Longer GCT compared to side-step • Athletes must time amplitude jump in evasive situations • Slowest strategy for pre-planned COD speed vs. side-step	• Attacking agility – 1 vs. 1 situations in sport to get past an opponent or to get into space • Evasive manoeuvres in sport to feint and deceive an opponent e.g., rugby, American football, soccer, etc. • Could be an effective strategy during situations with low approach velocity as small amplitude jump prior to push-off will engage SSC and subsequently increase lateral propulsion

Key: XOC: Crossover cut; COD: Change of direction; GCT: Ground contact time; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; SSC: Stretch-shortening cycle; ACL: Anterior cruciate ligament; ML: Medio-lateral

2.3.2 The role of the penultimate foot contact

A narrative review regarding the role of the PFC on COD with technical guidelines and braking strategy technique programme guidelines has been published (147) (Appendix 8.1), and the implications of PFC braking are discussed in Chapter 2.3.3 and 2.3.4.

2.3.3 Side-stepping biomechanical determinants of injury risk and performance: a performance-injury conflict?

Side-step cutting is the most commonly performed cutting action in multidirectional sports, including rugby (601, 632) and netball (184), and generally the "highest-risk" cutting action due to the highest reported incidences of non-contact ACL injuries observed performing this action (94, 376). As such, side-step cutting is the primary focus for this thesis. Insight into the mechanics and techniques associated with increased knee joint loading during COD enables a better understanding into the potential "high-risk" postures associated with non-contact ACL injury risk and can improve the screening and mitigation strategies to reduce ACL injury risk during directional changes. Consequently, a literature search was performed using *Medline* and *Sport Discus* databases, while additional searches of Google scholar were performed to understand the technical and mechanical determinants of knee joint loads during side-step cutting.

Search terms were as follows: 1) "biomechanics", or "knee joint load", or "knee abduction moment", or "ACL load", or "kinetics", AND 2) "change of direction", or "cutting", or "cut", or "sidestep", "side-step" or "run-and-cut". Bibliographies of potentially relevant studies were hand searched to identify any additional studies and citation tracking on Google Scholar was used to identify any additional material. The final search date was 19th July 2019. Articles were included for the review if: 1) they examined a pre-planned or unplanned COD that contained an approach run/sprint and a side-step cutting action; 2) conducted 3D motion and GRF analysis and evaluated KAM and/or KIRM; and 3) included healthy, injury free participants who participated in sport or physical activity. Studies that failed to meet the abovementioned criteria were subsequently excluded.

Most COD biomechanical associative injury risk factors studies have investigated KAMs during side-steps of 30-90° (51, 52, 67, 77, 94, 98, 99, 114, 133, 181, 182, 209, 218) while only a few have investigated KIRMs (50, 51, 67, 94, 209). Prospective research has found KAMs during a DVJ to predict non-contact ACL injury in female adolescent athletes (81), and KIRMs generally display greater variability and inaccuracies (370, 497) which most likely explains the popularity in assessing KAMs during side-stepping. However, it should be noted that a combination of knee abduction and internal rotation moments increases ACL strain to a greater extent than solely abduction, anterior tibial shear, or internal rotation moments (9, 107, 126, 152, 178), which makes it surprising that a paucity of research has investigated KIRMs.

A side-stepping deterministic model has been developed and adapted from Jones (276), outlining several technical, kinematic, and kinetic factors linked to generating greater KAMs (Figure 2.7 and Table 2.4). These identified "high-risk" parameters potentially predispose athletes to generating higher magnitudes of KAMs by increasing the moment arm, increasing the GRF, or a combination of the two (301). Consequently, this deterministic model can be used as a screening tool to evaluate poor movement quality during side-stepping, and can also be used as a technical framework for coaching safer side-stepping mechanics (278). A summary of research is provided below outlining how the trunk, hip, knee, ankle, foot, and braking strategy biomechanics influence side-step cutting knee joint loads, along with a biomechanical rationale outlining how the identified variable evokes potentially hazardous knee joint loading.

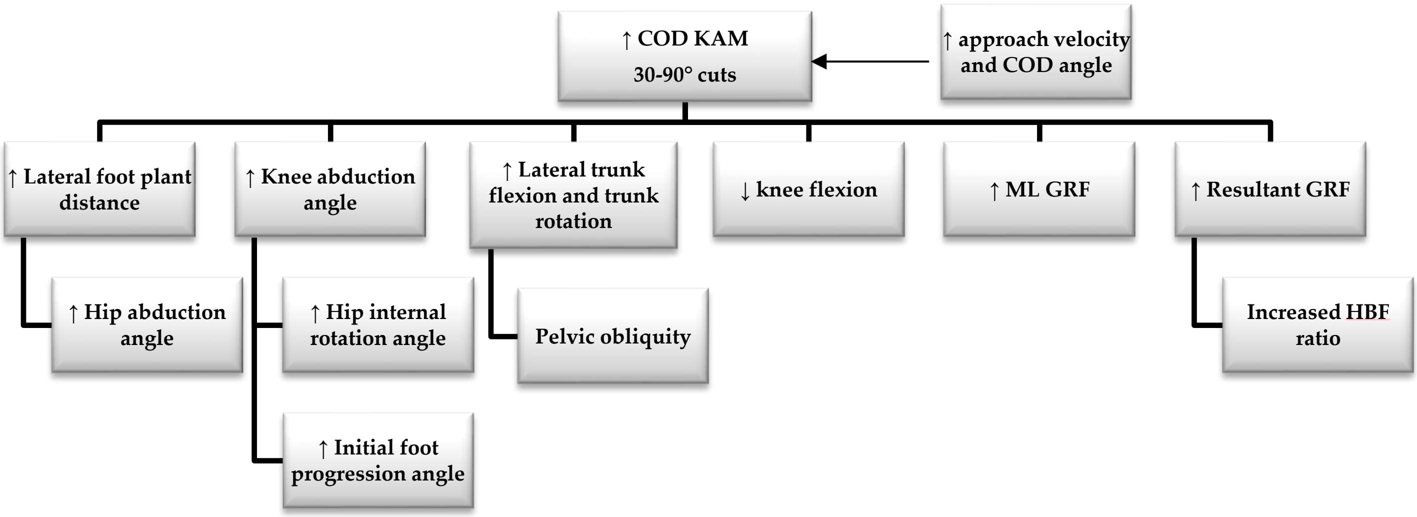


Figure 2.7. KAM deterministic model during side-stepping. Adapted from Jones (276) (Dempsey et al. (126, 127); Havens and Sigward (228); Jones et al. (272); Kristianslund et al. (301); Sigward and Powers (519); McClean et al. (358); Jamison et al. (267); Frank et al. (189); Donnelly et al. (141); David et al. (117); Staynor et al. (534); Weir et al. (593))

Table 2.4. Determinants of KAM during side-step cutting

Determinant of KAM	Association
Lateral foot plant distance	Jones et al. (272): (r = 0.45, r^2 = 20%, p < 0.05); Dempsey et al. (127): wide vs. narrow greater KAM (p ≤ 0.003, ES = 0.75-0.97) and greater KIRM (p = 0.048, ES = 0.79); Kristianslund et al. (301): ↑ 3.7° (cut width) = ↑ KAM by 17%; Havens & Sigward (228): (r = 0.472, r^2 = 0.22, p < 0.05)
Hip abduction angle	Sigward et al. (519): "High-risk" displayed > (p = 0.002, ES = 0.79); Weir et al. (593): Significant independent predictor (β = 0.012, SE = 0.006, p < 0.037)
Initial knee abduction angle	Jones et al. (272): (r = -0.67, r^2 = 45%, p < 0.01); Kristianslund et al. (301): ↓ of 4.4° = 19% ↓ in KAM; McLean et al. (358): males (r^2 = 0.21), females (r^2 = 0.35)
Initial foot progression angle	Sigward et al. (519): "High-risk" displayed > (p = 0.04, ES = 0.55)
Initial hip internal rotation angle	Sigward et al. (519): "High-risk" displayed > (p = 0.008, ES = 0.71); Havens & Sigward (228): (r^2 = 25%, p < 0.05); McLean et al. (358): males (r^2 = 0.56), female (r^2 = 0.60); Sigward et al. (516): significant predictor of KAMs

Lateral trunk flexion or trunk rotation	Jones et al. (272): (r = -0.42, r^2 =18%, p < 0.05) - IC; Dempsey et al. (127): trunk rotation (p = 0.030, ES = 0.50) - IC; Jamison et al. (267): torso angle (outside tilt = (p = 0.02) and torso GRF shoulder angle (p = 0.036)) – linear model compound symmetry covariance; Frank et al. (189): Trunk rotation displacement (r = -0.46, p = 0.011); Weir et al. (593): Significant independent predictor (β = 0.021, SE = 0.007, p < 0.001) - peak
GRF	Jones et al. (272): Medial GRF (r = 0.59, r^2 = 35%, p = .001); Sigward et al. (519): "High-risk" > laterally directed GRF (p < 0.001, ES = 1.51) (r = 0.61, p < 0.001); Jones et al. (273): PFC average horizontal GRF (r = -0.569, r^2 = 32%, p = 0.006); Sigward et al. (516): Vertical GRF CUT45˚ (r = 0.607, r^2 = 0.37%, p < 0.001), CUT110° posterior GRF (r = 0.460, r^2 = 19%, p = 0.001)
Rearfoot landings	Kristianslund et al. (301): feet pointing 16° more downward = 13% ↓ KAM; Donnelly et al. (139): Habitual RF greater KAM (p = 0.001, ES = 1.99); David et al. (117): greater KAM over 11-19% (p = 0.08) of stance phase.
Initial knee flexion angle	Weir et al. (593): Significant independent predictor (β = -0.025, SE = 0.006, p < 0.001)

Key: GRF: Ground reaction force; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; RF: Rearfoot; ES = Effect size; PFC: Penultimate foot contact; IC: Initial contact; SE: Standard error; β – parameter coefficients

Trunk biomechanics

Trunk neuromuscular control and positioning is a contributing factor to potentially hazardous knee joint loading (126, 127, 141, 189, 248, 266, 267, 272, 366, 636), with prospective studies showing deficits in trunk control and proprioception associated with increased knee and ACL injury risk (635, 636). With this in mind, several studies have examined the relationship between trunk control and knee joint loading during side-stepping (126, 127, 189, 267, 272, 593). Dempsey et al. (127) found 45˚ sidestep techniques which involved a wide lateral foot plant combined with the torso leaning in the opposite direction to the cut (lateral trunk flexion) increased KAMs compared to narrower cuts with upright trunk postures. In addition, the authors also found wide foot plants with the torso rotated in the opposite direction (over stance limb) resulted in significantly greater peak KIRMs (Table 2.4). Corroborating the findings of Dempsey et al. (127), Frank et al. (189) demonstrated trunk rotation displacement towards the stance limb increased frontal plane knee loading (Table 2.4), while Weir et al. (593) also observed peak lateral trunk flexion angles were significant predictors of KAMs. Similarly, Jamison et al. (267) also reported strong relationships between lateral trunk flexion and KAMs and KIRMs during a 45° side-stepping task (Table 2.4). Furthermore, during a greater angled 90˚ cutting task, lateral trunk flexion was also significantly related to KAMs (272) (Table 2.4), while lateral trunk flexion is also amplified during unanticipated tasks compared to pre-planned movements (381, 534).

Laterally flexing the trunk (deviation of the trunk) or rotating the trunk towards the plant leg shifts weight laterally, creating a more laterally directed force vector, thereby increasing its moment arm relative to the knee joint centre, and therefore increasing the resultant KAM (126, 127, 189, 267, 272). Conversely, trunk lean and rotation towards the intended direction of travel shifts the GRF vector more in line with the knee joint centre, thus decreasing the moment arm distance (141, 189). During the WA of a side-step cut, the entire body's mass must be balanced and supported on one leg. As the trunk contains over half of the body's mass, deficits in neuromuscular control and suboptimal trunk motion and position can be a critical factor in knee joint loading (248, 366). Visual observations of non-contact ACL injuries have commonly identified lateral trunk flexion and rotation towards the plant limb (injured limb) during CODs and landings (248, 541), which supports the findings of 3D biomechanical investigation that have shown the potential to generate potentially hazardous knee joint loading from poor trunk alignment during directional changes (126, 127, 267, 272).

Collectively, these findings suggest side-step techniques which encourage trunk lean and rotation towards the intended direction of travel may potentially reduce knee joint loading and subsequent risk of injury (126, 127, 189, 267, 272). Computer simulations of side-stepping have showed a reduction in knee valgus loading through shifting whole-body COM more medially (141), while COD technique modification interventions which focus on reducing lateral foot plant distances and maintaining an erect (frontal plane) trunk posture was found to reduce KAMs (126). Staynor et al. (534) has demonstrated greater pelvic obliquity towards the intended COD allows the trunk to remain more upright (i.e., reduced lateral trunk flexion over the plant limb) and argues that the pelvis acts as a "gimble"; playing a crucial role in trunk stability and alignment. Furthermore, from a performance perspective, trunk lean and rotation towards the intended direction of travel is also associated with faster COD performance (118, 228, 346); however, from an attacking agility (evasive) perspective, lateral trunk flexion may be a kinematic strategy employed by attackers to deceive opponents in order to evade and create separation distance from an opponent (61, 62, 252). Nevertheless, while attention should be placed on lower-limb alignment and motion, trunk positioning and motion should also be considered when screening and identifying potentially "high-risk" athletes. Emphasising upright and medially directed trunk positions appear to be suitable technique recommendations during side-stepping to reduce knee joint loading.

Hip biomechanics

While trunk kinematics can affect knee joint loading during directional changes, hip kinematics have also been shown to influence knee joint loading (228, 358, 516, 519, 593). Hip internal rotation angle has been identified as a potentially hazardous knee joint loading position with strong correlations observed between hip internal rotation angle and peak KAMs during a 45˚ side-step (358) (Table 2.4). This result is corroborated by previous research that also found hip internal rotation angle to be a significant predictor of KAMs during unanticipated 45˚ and 110˚ cuts (516). Additionally, research investigating a 90˚ cutting task found greater knee frontal plane loading was associated with an internally rotated hip (228) (Table 2.4). Furthermore, female subjects with greater KAMs compared to lower have also been shown to display greater hip internal rotation angles during a 45˚ side-step (519). Hip internal rotation angle is linked to KAA where biomechanically this posture leads to a more medially placed knee, resulting in a greater moment arm relative to the GRF vector, thus predisposing the athlete to larger KAMs (519). Therefore, based on the abovementioned evidence, internally rotated hip positions during side-stepping should be minimised to reduce the magnitude of KAMs, and thus, potential ACL injury risk.

Research from Sigward and Power (519) observed subjects with greater hip abduction angles demonstrated significantly greater KAMs (Table 2.4), while Weir et al. (593) has recently reported peak hip abduction angle as a significant predictor of peak KAMs during 45° side-step cutting. Biomechanically, greater hip abduction angles are connected to greater lateral foot plant distances (discussed in below section) because in order to create a wider lateral foot plant the hip must abduct to create such positions. Thus, this kinematic pattern (similar to lateral foot plant) moves the knee joint medial to the application of force creating a greater moment arm in the frontal plane and subsequently increasing the KAMs magnitude. Therefore, hip abduction angle and lateral foot plant distance could be modifiable technique factors to reduce KAMs during side-stepping.

Knee biomechanics

Biomechanical characteristics of the knee can have a significant effect on the strain experienced by the ACL, thus having a critical role regarding injury risk. Greater KAAs were reported in female athletes who sustained an ACL injury compared to uninjured athletes (242), while dynamic knee valgus is a commonly reported visual characteristic of non-contact ACL injuries during directional changes and landings (63, 94, 204, 270, 291, 298, 310, 426, 541, 583). Since the prospective study by Hewett et al. (242), several studies have explored the relationship between KAA and knee joint loading during side-step cutting (272, 301, 358, 516). McLean et al. (358) reported a strong association between initial KAA and KAMs during a 45° pre-planned side-step (Table 2.4). This finding is substantiated by previous research which has also reported a strong significant relationship between initial KAAs with KAMs during a sport-specific cut (301), pre-planned 90˚ cut (272), and unanticipated 45˚ and 110˚ cut (516) (Table 2.4). Greater initial KAAs and knee abduction motion during a COD can shift the knee medially relative to the GRF vector. Subsequently, this leads to a greater moment arm between the knee joint axis and the GRF vector, creating greater KAMs (272). Furthermore, McLean et al. (359) reported a change in KAA of 2˚ can lead to a 40 Nm change in KAM (assuming a GRF of 2500 N), thus greater risk of non-contact ACL injury (242). From a performance perspective, greater KAAs were not associated with faster 75° (346) or 45° and 90° cutting performance (228). Therefore, improving frontal plane knee control appears to be a viable strategy for reducing knee joint loading and subsequent ACL injury risk during side-stepping, with no associated performance detriments.

Extended knee postures with large anterior tibial shear can increase ACL strain (42, 43, 343, 615) and are also commonly observed visual characteristics of non-contact ACL injury (52, 63, 94, 204, 270, 291, 298, 310, 376, 426, 583). Moreover, stiffer landings and extended knee postures can increase GRFs and are associated with increased risk of ACL injury (321). Weir et al. (593) has recently reported smaller knee flexion angles at IC as a significant predictor of peak KAMs during 45° side-step cutting. As such, increasing knee flexion over WA during side-stepping could be a movement strategy to reduce knee joint loading; with active knee flexion advocated by Yu et al. (631). Dai et al. (115) found increasing knee flexion during 45° side-steps decreased posterior GRFs and KFMs, both of which have been associated with proximal anterior tibial shear (506, 631). However, it should be noted that the increased knee flexion technique concurrently increased GCT and resulted in lower exit velocities compared to natural side-stepping, and therefore negatively impacting COD performance. Similar to modulating foot plant distance during side-stepping, a "performance-injury conflict" is present regarding the role of knee flexion during side-stepping, whereby encouraging greater peak knee flexion and ROM could be detrimental to performance but decreases knee joint loading, and vice versa (183). Further research is required to confirm the chronic effect of knee flexion manipulation on COD knee joint loads and performance.

Foot and ankle biomechanics

During side-stepping actions, a lateral foot plant is required for generating ML propulsive impulse to accelerate into the intended direction of travel (11, 227, 261, 272) (Table 2.4); however, this action has the potential to evoke high knee joint loading during side-stepping (127, 228, 272, 301). For example, Jones et al. (272) found greater KAMs were observed with wider lateral foot plant distances during a pre-planned 90˚ cut. Similarly, this observation is also supported by previous research that have also shown greater lateral foot plant distance increases KAMs during sport-specific side-stepping (301) and a 45˚ cut (127) (Table 2.4). Although lateral foot plant distance was not quantified, Havens and Sigward (228) reported greater ML distances between COM and COP, indicative of lateral foot plant distance, were associated with greater frontal plane loading during a 45° cutting task (Table 2.4). A wide lateral foot plant position results in a more medial whole-body position with respect to the foot, creating a large GRF acting laterally outside the knee. Subsequently, this position results in a large moment arm between the perpendicular distance of the axis of rotation (knee) to the force (acting outside the knee), creating greater KAMs (228, 272, 519).

COD technique modifications that have coached narrower foot placements have been shown to reduce KAMs during pre-planned and unanticipated 45˚ sidestepping (126, 127). Thus, from an injury perspective, it would therefore appear intuitive to avoid wide lateral foot plants and coach narrower foot placements. Conversely, from a performance perspective, wider lateral foot plants are necessary to generate greater ML propulsive impulse (227, 261, 272) and are undoubtedly required for effective performance (228) and deceiving opponents (198). Subsequently, these findings suggest there could be a "performance-injury conflict" when executing cuts with wide lateral foot plants, which creates a predicament for practitioners who aim to minimise risk of injury while improving athletic performance. Arguably, bringing the foot closer to the midline could be detrimental to performance by limiting ML propulsive impulse, but no study to date has considered the implications for performance when narrowing lateral foot plant distance. If coaching narrower foot plant distances can reduce knee joint loading while maintaining performance, this could still be viewed as a positive and successful outcome. As such, further research is required investigating side-stepping technique modification, particularly lateral foot plant distance modifications on both injury risk factors and performance.

While the magnitude of lateral foot plant distance has been identified as a potential determinant of KAMs and been investigated extensively (127, 228, 272, 301), the orientation of the foot at IC during the COD can also influence knee joint loading (274, 519). Sigward and Powers (519) reported participants with greater KAMs displayed greater internally rotated foot progression angles (Table 2.4), while Dempsey et al. (126, 127) found a neutral position (i.e., closer to 0°) the safest foot placement strategy compared to internally and externally rotated positions. An internally rotated foot position at IC during DVJs has also been identified as a potentially hazardous position with greater KAAs, knee internal rotation angles, and KAMs reported (262, 561). This internally rotated foot posture appears to be suboptimal for the deceleration phase of COD because the large impact forces experienced during WA would be more likely to be attenuated in the frontal and transverse plane, potentially increasing KAMs (274). Conversely, a foot progression angle closer to 0˚ (straight) would allow the impact forces to be attenuated in the sagittal plane, utilising the large knee and hip extensor muscle groups (peak external knee and hip flexor moments) (274). Consequently, encouraging neutral foot positions during the IC and WA of side-steps appear to be a viable strategy for mitigating ACL injury risk.

The majority of literature has focused on hip, knee, and trunk kinetics and kinematics in relation to side-step knee joint loading; however, a recently emerging area of research is the impact of footfall pattern on COD biomechanics (106, 117, 139, 301, 625). Cortes et al. (106) compared rearfoot and forefoot contacts during an unanticipated 45° sidesteps in female soccer players, finding rearfoot techniques resulted in a more extended and abducted knee position, accompanied with greater KAMs. Yoshida et al. (625) found smaller KAAs during 60˚ side-stepping when performed with a forefoot strike pattern compared to rearfoot, which was accompanied with greater bicep femoris, semitendinosus, and lateral gastrocnemius pre- and post-contact activity (50 ms pre and post). While increasing hamstring activation is a positive outcome (341, 507, 591), the increased lateral gastrocnemius activity is of concern because the gastrocnemius is an antagonist to the ACL (2, 177, 522). Similarly, Donnelly et al. (139) found athletes demonstrating a habitual rearfoot technique in comparison to habitual forefoot during an unanticipated side-steps performed significantly more work and power through the knee joint. This was coupled with elevated non-sagittal plane peak ankle moments and greater peak knee flexor and abduction moments that may predispose athletes to a greater ACL loading (242, 342, 343, 512, 614). Furthermore, Donnelly et al. (139) found habitual forefoot strikers during side-stepping displayed greater peak ankle plantar-flexion moments compared to rearfoot. These findings are noteworthy because greater (internal) ankle plantar-flexor moments have also been associated with faster COD performance (228, 346).

Kristianslund et al. (301) also identified toe landing angle as a predictor of KAMs during a sport-specific side-step, stating a foot positioning 16˚ more downward would correspond to a 13% reduction in KAMs. Recently, David et al. (117) documented athletes who displayed rearfoot strike during a 90° cut generated greater KAMs, KIRMs, KFMs, and greater VGRFs compared to forefoot strikers. These findings are problematic because a combination of knee abduction and internal rotation moments increases strain on the ACL to a greater extent than uniplanar loading (26, 288, 342, 424, 513). A case study (68) of a non-contact ACL injury, using in-shoe pressure distribution sensors, during side-step cutting reported a heel strike posture as a characteristic of injury which was accompanied with greater medial and lateral heel GRF loading compared to the means of a CG. It has been argued that a heel strike is a provocative position which results in the propensity to generate high compressive and axial forces at the knee because the foot and lower-limb acts in a rigid, locked position as a single segment, resulting in a compromised attenuation of GRF at the ankle and greater reliance for the knee (53, 74). Conversely, a plantar flexed (forefoot) position is argued to be safer positioning by providing greater time to attenuate GRF and greater use of the ankle in energy absorption (53, 74).

Collectively, the forefoot contact could potentially be a safer side-step technique because rear foot techniques appear to evoke extended, abducted knee positions, and greater knee joint loads (106, 117, 139, 301, 625). These findings may partially explain the occurrence of non-contact ACL injury with heel strike foot contacts (54, 376). However, a large anterior placement of the foot relative to the COM is required to create posterior braking force, typically characterised with a rearfoot technique to facilitate braking (250). Additionally, it should be highlighted that a forefoot strategy increases (internal) ankle plantar-flexor moment and gastrocnemius activity, which acts as an antagonist to the ACL, and may contribute to ACL strain (tibial shear mechanism) (2, 177, 420). Forefoot techniques may also be more effective from a performance perspective due to the ability to create greater ankle plantar-flexion moments (228, 346). Thus, coaching forefoot side-stepping techniques could be a safer and faster strategy; however, further research is necessary to confirm the longitudinal effects of footfall technique modification on side-step injury risk and performance.

Ground Reaction Force and braking force ratio

The majority of the abovementioned associated "high-risk" mechanics have predominantly and theoretically increased the moment arm relative to the GRF (127, 228, 267, 272, 301, 358). However, as KAMs are a product of both the moment arm and GRF, increased GRFs during WA of directional changes are also a critical factor in knee joint loading (273, 516, 519). Sigward and Powers (519) found athletes who displayed greater KAMs demonstrated significantly greater laterally directed GRF (Table 2.4). The authors suggested a laterally directed GRF would impose a laterally directed intersegmental force at the distal tibia. As a result of its long lever arm (the perpendicular distance from the COM of the tibia to the distal end of the tibia), a larger laterally directed force would create a greater KAM. Likewise, Sigward et al. (516) found vertical and lateral GRFs, and resultant forces were predictors of KAMs for 45˚ cut, while posterior GRFs were predictors of 110˚ KAMs (Table 2.4). Conversely Jones et al. (272) found MLGRF was associated with greater KAMs during a 90˚ cut (Table 2.4); however, greater lateral foot plant distances were associated with greater MLGRF in order to generate ML impulse, but concerningly was also associated with greater KAMs. Thus, highlighting a conflict between performance and injury risk. Collectively, these findings highlight the potentially hazardous impact of greater GRFs during COD, thus reducing the magnitude of GRFs experienced during this phase could be an effective strategy in reducing knee joint loading; where most ACL injuries occur (52, 63, 94, 170, 298, 426, 583). Reducing FFC GRF could potentially be achieved by modifying athletes' braking strategies, coaching them to emphasise higher magnitudes and proportions of braking in the PFC relative to the FFC (147, 272-274), or manipulating lateral foot plant distance (126, 127, 272) or footfall.

The majority of COD biomechanical investigations have focused on the kinetics and kinematics of the FFC of the COD (127, 228, 267, 301, 358, 516, 519, 593). However, investigating the braking and deceleration characteristics of the PFC could provide greater understanding into optimal braking strategies to reduce hazardous knee joint loading (147, 272-274). Jones et al. (272, 274) investigated the peak HBFs during cutting and pivoting and although no significant relationships were observed, the authors found a trend in both studies that participants displaying greater KAMs had higher HBF ratios compared to the players displaying lower KAMs (-0.5 SD below mean) (CUT: 0.87 ± 0.04 vs. 0.82 ± 0.04; PIVOT; 0.99 ± 0.24 vs. 0.92 ± 0.18), respectively. More recently, Jones et al. (273) considered additional PFC braking characteristics, such as average force and impulse, and found athletes who demonstrated higher average HBFs in the PFC demonstrated lower KAMs during a 90˚ cut (Table 2.4). Collectively, these results are promising regarding the role of the braking characteristics of the PFC where it may be advantageous to demonstrate greater magnitudes and proportions of braking forces in the PFC relative to the FFC to reduce knee joint loading. However, further research in greater sample sizes is warranted, to confirm as to whether PFC dominant strategies are effective in reducing knee joint loading and facilitating faster performance during cutting.

Side-step cutting KAM determinants summary

Based on the associative work regarding determinants of KAMs during cutting, a deterministic model has been created identifying the variables which amplify KAMs, potentially predisposing athletes to an increased risk of non-contact ACL injury (Figure 2.7). These identified determinants influence the moment arm, GRF, or a combination of the two, therefore elevating KAMs; however, it appears that these biomechanical deficits are modifiable with appropriate training, feedback, and conditioning (93, 115, 125-127, 138, 140, 435). Particularly, technical characteristics associated with safer side-stepping are as follows: reduced lateral foot plant distances, thus lower hip abduction and orientating the foot closer to neutral; minimising knee valgus and hip internal rotation angles and motion at IC and WA; avoiding and limiting lateral trunk flexion and attempt to maintain an upright trunk position or trunk lean into the intended direction; reducing the magnitude of GRF during WA in the plant foot, potentially by attenuation through increased knee flexion and emphasising a greater proportion of braking in the PFC. The variables associated with increased KAMs support the commonly identified visual characteristics of ACL injuries including wide lateral foot plant with hip abduction, knee valgus, and the trunk flexed and/or rotated towards the plant foot (63, 94, 204, 248, 310, 426, 541, 583); thus, strengthening the argument to avoid and limit these potentially hazardous alignments, motions, and higher GRFs during rapid CODs. These determinants and biomechanical deficits can subsequently be used to identify athletes displaying "high-risk" patterns during screening and can also form the basis for ACL injury mitigation programmes.

Additionally, it should also be acknowledged that knee joint loading (sagittal, transverse, and frontal) is also increased during faster (114, 290, 301, 403, 574) and sharper (108, 212, 229, 505, 516) CODs. COD knee joint loading is also amplified in unanticipated situations (9, 38, 65, 98, 105, 126, 286, 289, 318, 437, 590), sport-specific cuts (79, 84, 174), and exacerbated under fatigue (98, 564, 591, 603); thus, it is imperative that athletes have the physical capacity (neuromuscular control, co-contraction, and rapid force production) and technique to tolerate the knee joint loading demands of side-steps (39, 272, 301, 328, 341, 410, 432, 544, 591), while every attempt should be made to ensure that athletes adopt optimal mechanics when performing directional changes (i.e., minimising knee valgus, lateral trunk flexion etc.)

2.3.4 Biomechanical determinants of change of direction speed performance

While directional changes have been extensively investigated from an injury risk perspective (126, 127, 189, 228, 267, 272-274, 301, 358, 515, 516, 519, 593), it is somewhat surprising that only a limited number of studies have comprehensively investigated the biomechanical determinants (i.e., 3D motion and GRF analysis) of COD speed performance (228, 346, 597), given the importance of COD manoeuvres in multidirectional sport. The following section will synthesise the current literature to date that has investigated the GRF, whole-body kinetic and kinematic, and technical determinants of COD speed performance.

<u>GRF determinants</u>

Several investigations have examined the GRF during the FFC of directional changes for an insight into the kinetic determinants of faster COD performance (143, 228, 338, 346, 530, 531, 596). Spiteri et al. (530) demonstrated stronger athletes ($n = 12$) in comparison to weaker ($n = 12$) athletes displayed significantly greater peak VPFs ($p = 0.012$; ES = 1.10), HBFs ($p = 0.004$; ES = 1.31), vertical braking impulse ($p = 0.028$; ES = 0.99), vertical propulsive impulse ($p = 0.025$; ES = 0.98), vertical total impulse ($p = 0.010$; ES = 1.15), greater horizontal braking impulse ($p = 0.004$; ES = 1.29), and horizontal total impulse ($p = 0.004$; ES = 1.70) during a 45˚ COD task resulting in faster COD exit velocity. Similarly, later work by Spiteri et al. (531) also found faster female basketball athletes ($n = 6$) compared to slower ($n = 6$) during the 505 produced significantly greater VBFs ($p = 0.02$, ES = 1.88) and VPFs ($p = 0.02$, ES = 1.72), while faster athletes during the T-test (90˚ cut) produced significantly shorter GCTs ($p = 0.001$, ES = 1.00), VBFs ($p = 0.01$, ES = 3.02), VPFs ($p = 0.001$, ES = 3.50), and vertical propulsive impulse ($p = 0.03$, ES = 0.91). No significant differences in approach velocities between faster and slower athletes were observed; therefore, the faster COD performance was attributed to the superior braking and propulsive forces and impulses, and spending less time braking and propelling, thus shorter GCT. This finding is supported by previous research that has also found shorter GCTs were associated with faster COD performance (143, 338, 346, 498, 531).

Supporting the abovementioned studies, researchers investigated a larger sample size ($n = 40$), and found faster male athletes during a modified 505 produced greater HPFs ($p \leq 0.002$, ES = 1.61-2.24) and shorter GCTs ($p = 0.077$-0.151,ES = 0.68-0.88) (143) highlighting the importance of high levels of propulsive forces in short GCTs (338, 346, 498, 531). However, in contrast to the results of Spiteri et al. (531), slower performance was associated with greater FFC VBFs ($r = 0.449$, $p < 0.01$), with faster athletes demonstrating significantly lower FFC VBFs ($p \leq 0.017$, ES = 1.19-1.46), suggesting a technical deficiency in braking force application by the slower athletes. Furthermore, faster athletes demonstrated significantly greater HPFs ($p \leq 0.002$, ES = 1.61-2.24) with no significant differences in VPFs ($p \geq 0.793$, ES $\leq$ 0.12). Similarly, Welch et al. (596) reported faster cutting performance was associated with greater magnitudes of force over 25 and 50 ms of ground contact, while greater horizontal to vertical braking and propulsive impulse ratios were associated with faster cutting performance in 25 Gaelic football players. Collectively, these studies highlight the importance of force vector specificity and the orientation of force application for effective reacceleration out of the COD. Biomechanical research into sprinting GRFs has demonstrated the importance of not only the magnitude of the resultant force, but the orientation of force application (378, 379). As force is a vector, possessing both magnitude and direction, several studies have demonstrated the importance of the magnitude of propulsive and braking forces (143, 530, 531) when changing direction, but there is a paucity of research examining the orientation of the braking and propulsive resultant forces, and providing the specific angles of these forces.

Theoretically, faster COD performance will also be influenced by the orientation of the braking forces to facilitate effective net deceleration, but also the orientation of propulsive forces will be integral for effective net reacceleration out of the COD. To date, Spiteri et al. (530) is the only COD study to examine the angle of peak HBF and HPF, reporting stronger athletes produced a significantly greater angle of HBF and HPF compared to weaker athletes during a 45° cut ($p = 0.001$, ES = 1.47-3.36); however, it was unknown if the orientation in force explained faster COD performance. Consequently, further research is required examining the magnitude and orientation of the braking and propulsive resultant forces during CODs for greater insight into optimal kinetic profiles for faster performance.

While the braking and propulsive GRF characteristics of the FFC have been identified as determinants of faster COD performance (143, 228, 530, 531, 596), PFC GRF braking characteristics are also an important component of faster performance (143, 202, 275). Preliminary work by Graham-Smith et al. (202) revealed faster 180° COD performance was associated with greater HBFs in the PFC (r = -0.674, r^2 = 45.4%, $p = 0.016$) and in the FFC (r = -0.579, r^2 = 33.5%, $p = 0.049$); however, the results were only presented in abstract format in a low sample size ($n = 12$). In a larger sample ($n = 40$), significant relationships between PFC peak HBF ($r = -0.337$, $p < 0.05$) and peak HBF ratio ($r = 0.429$, $p < 0.01$) with modified 505 performance, and faster athletes demonstrated greater PFC HBFs ($p = 0.027$, ES = 1.08) and lower HBF ratios ($p = 0.006$, ES = 1.50) compared to slower athletes have been observed (143). Moreover, Jones et al. (275) reported stronger female soccer players ($n = 9$) (eccentric knee extensor peak torque) demonstrated faster 505 performance and greater PFC peak and average horizontal GRFs (ES

= 1.00-1.23) compared to weaker players (n = 9). Collectively, the results of the aforementioned studies indicate a braking strategy which emphasises a greater proportion of braking forces in the PFC relative to the PFC, specifically in the posterior direction, could be effective in reducing horizontal momentum of the COM to allow more effective WA and preparation for the drive-off phase of directional changes (143, 269, 272, 275). However, the abovementioned studies are only representative of 180° tasks and as the biomechanical demands are angle dependent (39, 108, 212, 213, 227, 229, 504, 505, 516), evaluations of the PFC braking characteristics of different angled cuts and turns from a performance perspective warrant further investigation.

<u>Whole-body kinetic and kinematic determinants</u>

Currently, there are a limited number of studies that have examined the kinetic and kinematic determinants of faster COD speed performance (228, 346, 498, 530, 597). Sasaki et al. (498) investigated the relationship between trunk kinematics and completion time during a modified 505 in twelve college soccer players, reporting a significant positive correlation (r = 0.61, p < 0.05) between forward angular displacement of the trunk (between foot-contact and maximum trunk inclination) and completion time; highlighting the importance of trunk stability for effective COD. It should be noted, however, that several weaknesses exist in the study by Sasaki et al. (498) including the low sample size (n = 12) and only trunk kinematics were explored.

Conversely, Marshall et al. (346) conducted a comprehensive assessment of the whole-body kinetic and kinematics of 75° cutting performance in 15 Gaelic hurling players, finding five variables associated with faster performance (p < 0.01): peak ankle power (r = 0.77), peak ankle plantar-flexor moment (r = 0.65), range of pelvis lateral tilt (r = -0.54), maximum thorax lateral rotation angle (r = 0.51), and GCT (r = 0.48). Similarly, Welch et al. (597) using a novel technique of principal component analysis and permutation testing, found techniques with a low COM, a short GCT, maintenance of wide lateral foot plant, resisting hip flexion, rapid hip and knee extension, and trunk lean towards the intended direction of travel were associated with faster side-step (45° and 110°) cutting performance in 25 Gaelic football players. Additionally, David et al. (118) found faster 90° cut performers displayed earlier and greater pelvis pre-rotation during the PFC and FFC, resulting in shorter GCTs and higher exit average velocities. Collectively, these results suggest that techniques which encompass rapid levels of force production about the ankle, knee, and hip, maintenance of pelvic control, torso and pelvis lean and rotation towards the intended direction, a low COM, and a wide lateral foot plant, in short GCTs are effective for side-step cutting performance.

Havens and Sigward (228) explored the GRF and whole-body mechanics in 25 soccer players during 45° and 90° cutting tasks. Completion time for 45° cut performance was significantly correlated ($p < 0.05$) with hip power generation in the sagittal plane ($r = -0.475$) and peak ankle plantar-flexor moment ($r = 0.450$), which supports the results from Marshall et al. (346). These findings are unsurprising because concentric ankle power is pivotal for impulsive activities such as sprinting (123) and vertical jumping (572). Shorter 45˚ completion times were also associated with greater hip extensor moments ($r = 0.393$) and larger ML COM-COP distances ($r = -0.387$) (228). Additionally, greater ML impulse ($r = -0.489$, $p = 0.013$), hip rotation angle at IC ($r = -0.471$, $p = 0.018$), hip frontal power ($r = -0.586$, $p = 0.002$), and knee extensor moment ($r = 0.499$, $p = 0.024$) were correlated with faster 90° cut performance. These biomechanical factors demonstrate the importance of the lower-limb triple extension musculature and specifically highlight the importance of rapid force generation in the frontal plane for cutting performance due to the ML redirection requirements observed in cutting.

A key factor in ML propulsion during CODs is lateral foot plant distance (11, 228, 261, 272). Inaba et al. (261) compared nine different lateral foot plant distances during side-steps (distances 20-100% of height) observing greater hip extensor, knee extensor, and ankle plantar-flexor moments and joint works with wider lateral foot plants. In addition, the greater moments and joint work were accompanied with greater lateral component of velocity of COM and an increased lateral orientation of force vector. Additionally, Welch et al. (597) and Havens and Sigward (228) have both identified a wider lateral plant is associated with faster cutting performance. As such, these findings indicate that ML propulsion during side-step plant-and cut actions is influenced by the magnitude of foot plant distance due to its impact on the joint work and moment of the hip, knee and ankle, GRF magnitude and the orientation of the force vector. Therefore, when performing side-steps from a performance perspective, coaching a wide lateral foot plant is recommended because it facilitates greater velocities at push-off, thus faster performance.

<u>Importance of approach velocity</u>

In contrast to assessing the determinants of COD performance via 3D motion and GRF analysis, a novel method implementing laser speed guns to assess COM related kinematics has been used (213). The authors reported minimum speed reached during the COD was a large to very large determinant of 45˚ and 90˚ cut performance, while peak acceleration and peak speed additionally contributed. Likewise, Jones et al. (275) recently found approach velocity (at PFC) was moderately associated ($r = -0.484$, $r^2 = 23\%$) with faster completion times during a 180˚ COD speed task. The finding that faster approach velocities are a major determinant of COD speed performance is unsurprising because faster athletes will cover greater distances in less time. Therefore, the ability to maintain velocity or minimise the decline in velocity prior to- and during cuts of 45-90°, and shallow directional changes (≤ 45˚), is noteworthy and appears advantageous for faster COD performance. However, the ability to maintain velocity or minimise the decline in velocity in the COD will be dictated by the angle, approach distance, and subsequent entry velocity because larger braking forces and decelerations will be required during CODs of sharper angles (108, 227, 273, 516) and approach velocities (114, 574).

Determinants of COD performance summary

A deterministic model for cutting performance is presented below (Figure 2.8), indicating that velocity, creation of dynamic momentary instability, simultaneous rapid joint movements, and braking and propulsive impulse are integral for faster cutting performance. It should be noted that much of the published research into the kinetic and kinematic determinants of faster cutting performance have been low in sample size ranging from 12 to 25 (228, 346, 498, 530, 531), generally only investigated the plant foot (FFC) (228, 346, 498, 530, 531), and have not considered resultant force of the PFC and FFC. Thus, more comprehensive biomechanical investigations considering the aforementioned limitations are warranted to improve our understanding of the determinants and optimal techniques for effective COD performance, while considering the implications of these factors on knee joint loading.

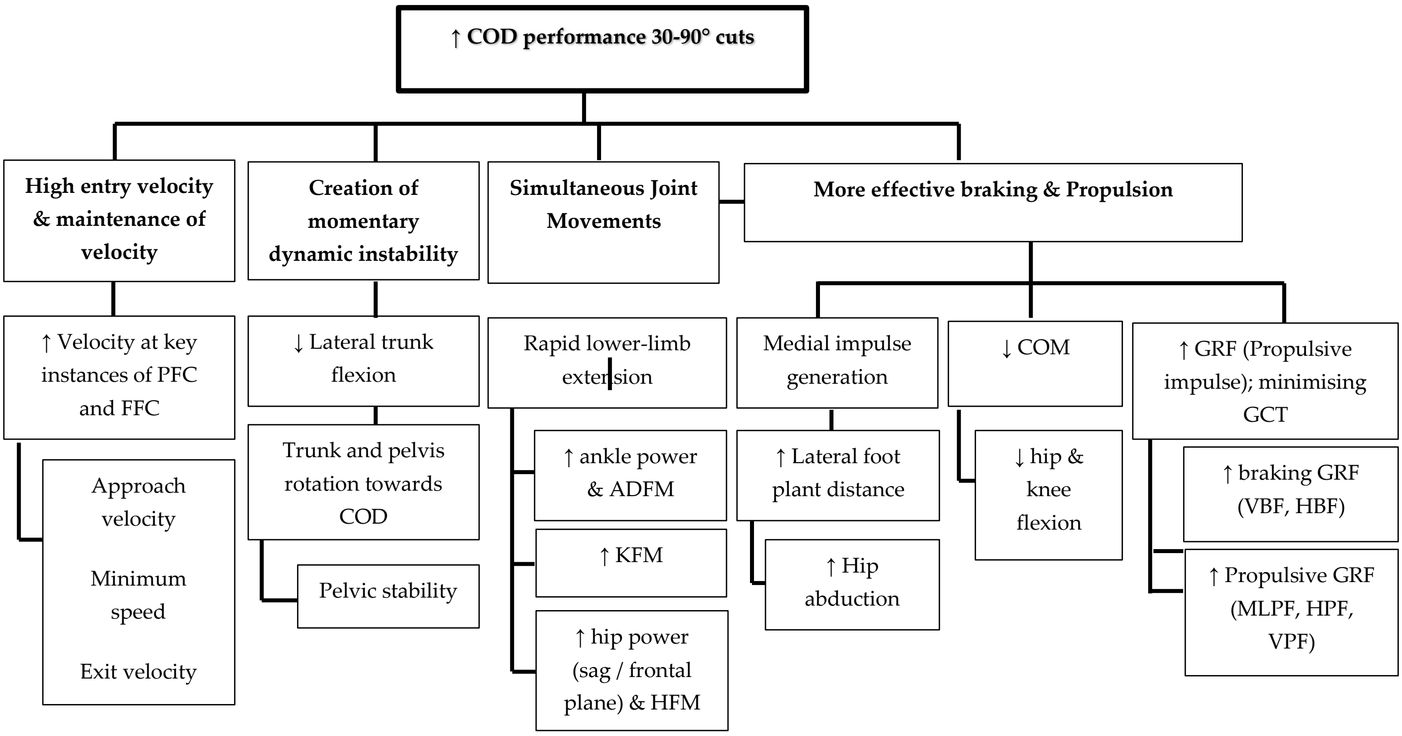


Figure 2.8. Cutting performance deterministic model during side-step cutting (Hader et al. (213); Inaba et al. (261); Jones et al. (272, 275); Havens and Sigward (228); Andrews et al. (11); Spiteri et al. (530, 531); Dos'Santos et al. (143, 146, 147); Graham-Smith et al. (202); Maloney et al. (338); Marshall et al. (346); Sasaki et al. (498); Welch et al. (596, 597))

2.3.5 The effect of training interventions on change of direction biomechanics associated with increased anterior cruciate ligament loading: a scoping review

(a proportion of this section has been accepted for publication in Sports Medicine (149) - Appendix 8.1)

In order to reduce ACL loading and potential injury risk during directional changes, particularly non-contact ACL injury, an effective strategy is to modify an athlete's movement mechanics by addressing biomechanical and neuromuscular deficits. This can be done through biomechanical and neuromuscular informed training interventions to reduce the magnitude of knee joint loading (183, 241, 245, 246, 327, 328, 391, 435, 508, 535). Due to the prevalence of non-contact ACL injuries associated with COD actions in multidirectional sport (63, 94, 270, 291, 298, 376, 426, 583), various training interventions have been performed in an attempt to alter COD biomechanical characteristics associated with increased ACL loading. These include COD technique modification training (126, 277), COD speed and footwork (606), mixed training programmes (sessions which integrates exercises from several training modalities i.e., plyometrics, stretching, balance, trunk stabilisation, and/or resistance training) (33, 535, 592, 594, 622), combined trunk stabilisation and resistance training (266), resistance training (93, 266), combined COD technique modification and balance training (140), combined resistance training and intersegmental control training during running and COD drills (292), dynamic core stability training (602), balance training (92, 93, 425), perturbation-enhanced plyometric training (599), and injury prevention warm-up protocols (i.e., Oslo, Core-Pac, F-MARC 11+) (33, 75, 76, 535, 555, 556, 638, 639). As practitioners working in multidirectional sports are interested in injury risk mitigation strategies, understanding the most effective training modalities that address COD biomechanics associated with increased ACL loading is of great importance. The purpose of this scoping review was three-fold: 1) to critically appraise and comprehensively synthesise the existing literature related to the effects of training interventions on COD biomechanics associated with increased knee joint loads and subsequent ACL loading; 2) to identify gaps in the literature and recommend areas for future research; and 3) to provide evidence-based recommendations which outline efficacious strategies for addressing COD biomechanics associated with increased ACL loading and potential non-contact injury risk.

2.3.5.1 LITERATURE SEARCH METHODOLOGY

A literature search was performed using *Medline* and *Sport Discus* databases. Figure 2.9 provides a schematic of the search methodology in accordance to PRISMA guidelines (367). Search terms were as follows: 1) "biomechanics", or "neuromuscular", or "electromyography", AND 2) "change of direction", or "cutting", or "cut", or "sidestep", or "turning", AND 3); "intervention", or "program", or "programme", or "training", or "modification". Bibliographies of relevant studies were hand searched to identify any additional studies. Citation tracking on Google Scholar was also used to identify any additional material. The search date ranged from August 15th, 2018-10th January, 2019. Articles were included for review if they met the following criteria:

1. Investigated a cutting or turning task (e.g., side-step, plant-and-cut actions, pivot)
2. Examined the effects of a training modality intervention (minimum 4 weeks) on COD biomechanics associated with increased ACL loading (e.g., knee valgus angle, knee abduction moments, knee flexion angle, knee rotation moments, knee flexion moment, vertical and posterior GRF, muscle activation, lateral trunk flexion, trunk rotation, foot progression angle etc.).
3. Included participants who participated in sport or physical activity.

Studies that failed to satisfy the abovementioned criteria were subsequently excluded. Training intervention studies that satisfied the abovementioned criteria were than classified into the following training modalities:

- Change of direction technique modification training: COD drills performed with coach feedback and cues that focus on modifying COD technique, such as lateral foot plant distance/ trunk positioning.
- COD speed and footwork training: pre-planned COD drills with no coach feedback or cues regarding COD technique.
- Balance training: balance training which incorporates stable and unstable training methods, such as balancing on one leg (while catching a ball), wobble boards etc.
- Mixed training programmes: session which integrates exercises from at least three or more of the following training modalities: plyometrics, stretching, balance, trunk stabilisation, and/or resistance training. These involve dedicated sessions performed outside typical sport-specific practice and games.
- Resistance training: session which includes free-weight and/or machine-based resistance training.
- Perturbation enhanced plyometrics: plyometric training performed with added perturbation (motorised platform) over WA.
- Trunk stabilisation training or dynamic core stability training: Trunk stabilisation training refers to training with static exercises (i.e., planks etc). Dynamic core stability training includes exercises performed dynamically (i.e., dynamic planks, bridges etc) with added perturbations.

- Combined training: training which combines two of the abovementioned training modalities. These are sub-divided into: combined COD technique modification and balance training, combined trunk stabilisation and resistance training, and combined resistance training and intersegmental control training during running and COD drills.
- Warm-up interventions: Neuromuscular warm-up interventions that were typically 15-25 minutes performed prior to sport-specific practice (i.e., technical and tactical) and / or games. These warm-ups replaced their normal skill/tactical warm-up, and include exercises from various training modalities, such as trunk stabilisation, plyometrics, balance, body weight resistance training, running and COD drills. These include the Oslo Neuromuscular warm-up, Core Position and Control (Core-Pac) warm-up intervention, and FIFA's Medical Assessment and Research Centre 11+ (F-MARC 11+) soccer-specific injury prevention warm-up.

The following sections outline the findings of included studies relevant to the effects of specific training interventions on COD biomechanics associated with ACL loading.

2.3.5.2 RESULTS

Initial database searches resulted in the identification of 1021 articles, with an additional 6 articles through bibliographies, citation tracking, and hand searching (Figure 2.9). After removing duplicates, 928 articles were retained for initial screening. Title and abstract screening resulted in 889 articles excluded. The remaining 29 articles were further examined using the inclusion/exclusion criteria and 7 studies were excluded, resulting in 25 datasets from 22 studies included to examine the effect of training interventions on COD biomechanics associated with increased ACL loading (Figure 2.9 and Tables 2.5-2.8). Eleven of the 22 studies failed to include a CG (Tables 2.5-2.9). Only one study provided reliability measures for biomechanical variables, but no study acknowledged measurement error or established smallest worthwhile change (SWC) or SDD when interpreting findings (Tables 2.5-2.8). The effects of these training interventions on COD biomechanics are presented in Tables 2.5-2.8.

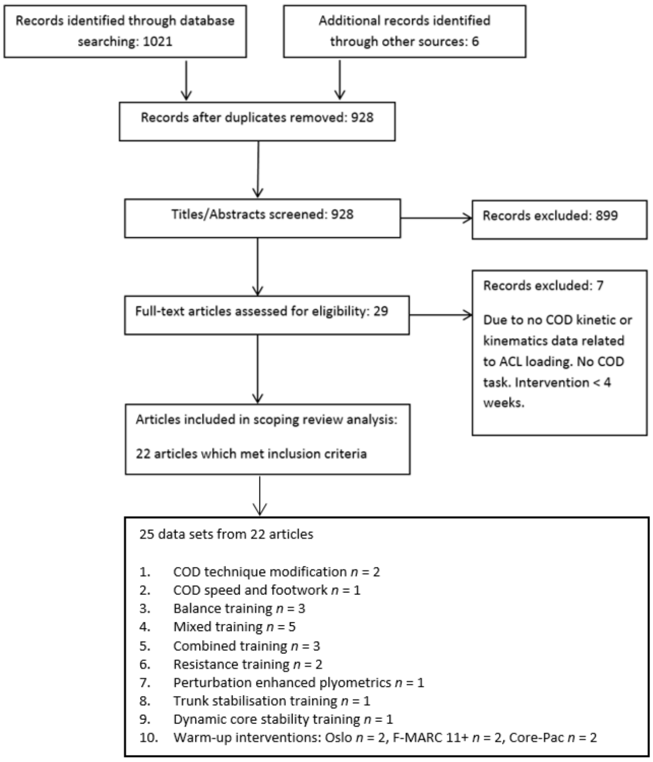


Figure 2.9. Flow diagram illustrating the different phases of the scoping review; based on PRISMA recommendations. COD: Change of direction; ACL: Anterior cruciate ligament

2.3.5.3 DISCUSSION

The primary purpose of this scoping review was to critically appraise and comprehensively synthesise the existing literature related to the effects of training interventions on COD biomechanics associated with increased knee joint loads and subsequent ACL loading, and identify gaps in the literature with subsequent recommended areas for further research. The primary findings were balance and COD technique modification training appear to be the most effective training modalities for reducing knee joint loading (small to moderate effect sizes [ES]) during COD while other training modalities were generally ineffective (Tables 2.5-2.8). Although the published literature regarding the effectiveness of training interventions on COD biomechanics associated with increased ACL loading is indeed insightful, there are key methodological and research design limitations which must be acknowledged going forward to improve our understanding of effective training strategies that reduce COD knee joint loads. These limitations include the lack of CGs (eleven studies contained no CG), failure to establish reliability measures (21 studies) and acknowledging measurement error to establish "real" and "meaningful" changes, and generally failing to consider the implications on performance. Additionally, changes in biomechanics are based on mean group differences, with little consideration for individual responses. The effectiveness of the different training modalities, gaps in the literature, and recommended areas of further research are discussed by training modality below.

COD technique modification training

In order to reduce knee joint moments and subsequent ACL loading, the magnitude of the GRF or the moment arm must be reduced (301). Several studies have shown that acute (within-session) changes in COD technique can reduce knee joint loads (76, 115, 127), such as narrowing lateral foot plant distance and changing trunk orientation (127), increasing knee flexion (115), and moving the COM closer to the base of support (76). Because of the promising results observed with acute COD technique modification, several studies have investigated the chronic effects of COD technique modification on COD biomechanics associated with increased ACL loading (126, 277) (Table 2.5).

Dempsey et al. (127) initially examined the effects of acute, within-session COD technique modification (altering foot plant distances, trunk positioning, and foot orientations) on 45° side-step biomechanics. A wide foot plant combined with lateral trunk flexion over the plant foot resulted in the greatest peak KAMs ($p \leq 0.003$, ES = 0.75-0.97), while a wide lateral foot plant with torso rotation towards the plant leg resulted in significantly ($p = 0.001$, ES = 1.00) greater peak KIRMs. Conversely, a side-step technique which involved neutral foot positioning, a foot plant distance closer to the midline, and an upright (in frontal plane) torso resulted in the lowest knee joint loading (KAM and KIRM), due to reducing the moment arm between the GRF and knee joint centre (127). As such, a narrow foot placement with an upright trunk was subsequently advocated as a safer side-stepping technique (127).

Expanding on the promising results of the acute side-stepping technical modification, Dempsey et al. (126) investigated the effects of a 45° side-stepping technique modification intervention over 6 weeks (2 × 15 mins sessions per week) on COD biomechanics (Table 2.5). The intervention consisted of performing side-step drills with imposed technique changes by bringing the foot closer to the midline (tape placed on floor for acceptable foot plant distance), maintaining an upright torso, and torso facing towards the direction of travel. Importantly, participants were provided with oral and video feedback regarding their technique between repetitions. The authors, notably, demonstrated significantly lower peak KAMs ($p = 0.034$, ES = 0.58-0.78, 36%) during both anticipated and unanticipated side-step tasks accompanied with significant reductions in lateral foot plant distance and lateral trunk flexion ($p \leq 0.039$, ES = 0.14-1.09) (Table 2.5). As such, side-step technical modifications were effective in reducing knee joint loading, and in turn, could be an effective strategy to reduce non-contact ACL injury risk.

Although the acute (127) and chronic COD technique modifications (126) by Dempsey et al. have shown positive reductions in knee joint loading during directional changes, a note of caution is warranted. Firstly, the abovementioned studies have failed to present and acknowledge measurement error values; thus, it is uncertain whether such changes were greater than the measurement error, and therefore "real". Secondly, the training intervention performed by Dempsey et al. (126) did not contain a CG; therefore, the results should be interpreted with caution. Although reducing lateral foot plant distance was shown to reduce peak KAMs (126), critically, this imposed technique change could be detrimental for ML force application and may result in suboptimal COD performance (i.e., reduced exit velocity from the push-off) (227, 228, 261, 272). It is worth noting, however, athletes adopted less lateral trunk flexion (i.e., more upright trunk) which may be a positive adaptation for faster cutting performance (346). Moreover, the studies performed by Dempsey et al. (126, 127) have failed to consider the implications of such changes in side-step technique on COD performance (i.e., GCT, COD exit velocity, and completion time). As athletes are driven by performance, they may be unlikely to adopt movement techniques which decrease risk of knee injury if they do not result in effective performance (228). Consequently, further research is necessary investigating the chronic effects of side-stepping technique modification on both biomechanics associated with decreased ACL loading and increased performance (183).

Investigating a sharper COD (180°), Jones et al. (277) reported a reduction in turning KAMs (ES = 0.73) and improved completion times (ES = 0.74) in female netball players as a result of a six-week COD technique modification intervention which consisted of technical drills that encouraged PFC braking, backwards trunk inclination, and neutral foot positioning (Table 2.5). Interestingly, a strong association between changes in IFPAs and KAMs (r^2 = 37%, p = 0.028) was observed, while athletes also demonstrated changes in trunk inclination during the FFC (ES = 0.58). However, similar to Dempsey et al. (126), there was no CG, and findings were not interpreted in relation to the measurement error. Nevertheless, instructing athletes to adopt a more neutral foot progression angle (i.e., closer to 0°) during sharper 180° turns could be an effective strategy to reduce peak KAMs and subsequent ACL loading.

Collectively, COD technique modification appears to be a potentially viable and an effective strategy in reducing knee joint loading (Table 2.5); however, published COD technique training interventions lack CGs and do not acknowledge measurement error when interpreting findings. Moreover, it is unknown how long such changes in COD biomechanics are retained for following a training intervention. Thus, further COD technique modification interventions are required that include a CG and acknowledge measurement errors to definitively confirm the effectiveness of this training modality in reducing knee joint loading. Moreover, COD performance should also be considered to understand implications of such technical modifications on knee joint loading and performance because athletes may be unlikely to adopt safer strategies at the expense of performance. If COD performance can be maintained or improved while simultaneously reducing knee joint loading following COD technical drills modification, this would help improve adherence and may provide practitioners an effective strategy to mitigate injury risk.

COD speed and footwork training

Wilderman et al. (606) examined the effects of a 6-week agility training programme which was performed four times a week by female basketball players compared to a CG. The programme consisted of pre-planned COD speed, footwork, and manoeuvrability drills; thus, the term "agility" is incorrect due to the absence of drills that involve responding to an external stimulus (131, 628). Nevertheless, the intervention group (IG) showed increases in medial hamstring activation (ES = 0.94) (Table 2.5) which may help reduce anterior tibial shear and subsequent ACL strain (249, 322, 341, 396, 496), though no statistically significant (ES ≤ 0.15) changes in KFA or VGRFs were observed. A limitation of this study was the lack of specific drills that focused on side-stepping mechanics. In addition, the absence of coach feedback regarding the athlete's technique is also a limitation that may explain the mixed results. Conversely, studies that have documented positive changes in COD technique (126, 277) have emphasised the importance of coach technical feedback. It is also worth noting that the biomechanical variables examined during the side-step by Wilderman et al. (606) were limited to only KFA, GRF, and muscle activity; thus, a more comprehensive analysis of frontal plane biomechanics and trunk kinematics would have strengthened this study, because these factors are strong determinants of knee joint loading (127, 183, 301).

Balance training

Because lower-limb balance training has been shown to be effective in reducing ACL injury rates in sport (72, 398), several studies have attempted to identify the underlying biomechanical and neuromuscular mechanisms which may explain the reductions in ACL rates (Table 2.6). Oliveira et al. (425) demonstrated 6-weeks of balance training resulted in a statistically significant 33% reduction in peak KAMs during a perturbed cutting task, while a CG demonstrated a slight increase, though not statistically significant (Table 2.6). The improvement in peak KAMs was accompanied with increased electromyography (EMG) activation of the trunk and proximal hip musculature and increased EMG burst duration prior to IC (Table 2.6). Although trunk kinematics were not examined, the authors hypothesised the improved muscle activity of the hip and trunk lead to improvements in trunk control, which is a critical factor for knee joint loading (127, 189, 267, 272). It is worth noting, however, that pre- and post-analysis in perturbed cutting biomechanics and muscle activation was only performed for one trial. This is a problematic issue because evaluations based only on one trial can lead to invalid data and erroneous conclusions (23, 264), while one trial may not be fully representative of an athlete's typical movement pattern (264).

Reporting a similar finding to Oliveira et al. (425), but investigating a greater trial size, Cochrane et al. (93) found balance training was the most effective modality to reduce both peak KAMs ($p < 0.001$, 62%) and peak KIRMs ($p < 0.001$, 32%) in all anticipated and unanticipated COD manoeuvres (Table 2.6), compared to machine-based resistance training, free weight, and combined machine-based and balance training. While machine-based training was also effective in reducing peak KAMs ($p < 0.05$, 27%), free weight and combined machine-based weights and balance training were ineffective in reducing KAMs or KIRMs (Table 2.6), and a CG increased their peak KAM. The reductions in frontal and transverse plane joint loads as a result of balance training may be explained by earlier work from Cochrane et al. (92) that found 12-weeks balance training to elicit positive and potentially safer changes in lower-limb muscle activation. Increased knee flexor/extensor contraction ratios, increased flexor muscle activation, and increased biceps femoris/semimembranosus contraction ratios were observed, while a strength training group increased their quadriceps activation and reduced their hamstring activation (Table 2.6). The hamstrings are considered to have an important role during the WA phase of COD in preventing anterior tibial translation and reducing anterior tibial shear and ACL strain (249, 322, 328, 341, 396, 496).

Consequently, the results from these studies suggest that balance training could be an effective training modality for reducing COD knee joint loading (Table 2.6) and subsequent ACL loading. The successful results are most likely attributed to positive changes in hamstring, hip, and trunk muscle activation which supports and reduces knee joint loading (328). It is worth noting, however, that the aforementioned studies have failed to acknowledge measurement error when interpreting their findings and not considered the performance implications; and thus, is a future direction of research to definitively conclude the effectiveness of this method. Nevertheless, balance training involves the use of wobble boards, instability surfaces, and catching a ball which is easy to perform, simple to regress and progress, and can be easily integrated into athletes training programmes to help reduce ACL loading and potential injury risk.

Mixed training programmes

Several studies have used mixed training programmes (sessions which integrates exercises from several training modalities i.e., plyometrics, stretching, balance, trunk stabilisation, and/or resistance training) (33, 535, 592, 594, 622) or a combinations of training modalities in an attempt to alter COD biomechanics associated with increased ACL loading (Table 2.7).

Combination of balance and COD technique modification training

Based on the successful results of previous balance (93) and COD technique modification (126) interventions, Donnelly et al. (140) inspected the combined effects of balance training and COD technique modification compared to acceleration training on COD biomechanics. This intervention was performed in Australian Rules footballers (1001 male athletes) over a regular season in a "real-world" environment. Both training groups reduced their peak KIRM during pre-planned side-steps (45% reduction), but peak KAMs significantly increased during unanticipated side-steps (31% increase) following the training intervention (Table 2.7), failing to substantiate the positive findings of previous research (93, 126). Similar to previous COD technique modification and balance training interventions (Tables 2.5 & 2.6), changes in knee joint loads were not interpreted in relation to the measurement error. The mixed findings of the training intervention by Donnelly et al. (140) could be explained by the low compliance rate of only 45% reported for the training intervention and a high athlete to coach ratio (40:1). These issues are problematic because successful training interventions that reduce knee joint load, thus ACL loading, are fundamentally underpinned by compliance (82, 93, 126, 241, 323, 432, 435). Furthermore, the high athlete to coach ratios prevent sufficient biomechanical technique correction and feedback to individuals, which again limits the effectiveness of technique modification interventions (82, 126, 241, 244, 323, 432, 435). Additionally, only a subset of 34 athletes were examined for biomechanical testing throughout the season; thus, it is uncertain whether the subset's biomechanics are fully representative of the whole sample (n = 1001).

Although balance (93) and COD technique modification (126) has been shown to be effective in reducing knee joint loading in controlled environments, and in relatively small sample sizes (Table 2.5 & 2.6), the study by Donnelly et al. (140) highlights the potential difficulty in administering such training methods in "real-world" environments at the community-level. The low adherence may be evident in such strategies to community-level athletes, who may not have the time or desire to complete further training outside typical sports practice, while the high athlete to coach ratio often associated at the amateur and community-level makes it potentially unrealistic to provide individualised feedback. Therefore, these issues present a potential barrier in applying such strategies in the "real-world" to attempt to reduce injury risk or investigate injury risk. Nevertheless, based on these findings, in order to perform a successful technique intervention which reduces knee joint loading, thus relative risk of injury, it is essential that there is high compliance and individual feedback regarding the athlete's technique to facilitate effective changes in COD biomechanics (126, 432, 435).

Combination of trunk stabilisation and resistance training

Jamison et al. (266) compared the effects of combined resistance and trunk stabilisation (static trunk exercises) training compared to resistance training only on trunk control, strength, and knee joint loading during a 45° unanticipated side-step. Significantly greater peak KAMs ($p = 0.012$, 50%) were observed for the resistance training group only, and although not statistically different ($p = 0.116$), the combined group also displayed a 35% increase in side-stepping KAMs (Table 2.7). Conversely, the combined group demonstrated a 35% reduction in peak KIRMs, though this was not statistically significant ($p = 0.110$), whereas KIRMs increased 12% in the resistance training group ($p = 0.617$) (Table 2.7), though these changes were not interpreted in relation to the measurement error. Unsurprisingly, the combined group showed significantly greater improvements in core endurance and strength, while both groups improved maximum deadlift strength (Table 2.7). This finding is similar to Cochrane et al. (92, 93) who also found resistance training was ineffective in reducing peak KAMs during COD tasks, potentially due to the reduced hamstring and increased quadriceps activation which may contribute to increased knee joint loads (328). Although performance measures were not examined (i.e., completion time, exit velocity, GCT) in the studies by Jamison et al. (266) and Cochrane et al. (93), the groups which performed resistance training increased their strength. Thus, it is speculated that the increased peak KAMs could be a by-product of an increase in approach velocity and an increased ability to produce force due to the strength training; both of which can influence knee joint loading (148, 183, 574).

Collectively, resistance training and combined resistance training and trunk stabilisation modalities appear ineffective in reducing COD knee joint loading (Table 2.7). The ineffectiveness of these training modalities, however, could be explained by the lack of task-specific training around trunk and lower-limb control associated with multiplanar side-stepping (126, 599, 602). Additionally, it should be noted that the trunk stabilisation intervention only included static exercises; however, dynamic trunk stabilisation exercises with perturbations may have provided a greater stimulus and specificity in order to reduce side-stepping knee joint loading (599, 602). Furthermore, it is also worth acknowledging that a low sample size was investigated in the study by Jamison et al. (266) ($n = 10$ and 11) which failed to achieve adequate statistical power (*a priori* determined minimum sample of 18). It is must be noted, however, that although resistance training does not reduce knee joint loads during COD (Tables 2.6 & 2.7), resistance training provides several benefits for athletes including enhanced performance during dynamic tasks (i.e., jumping, sprinting, COD) and positive adaptations to tissues (muscle, bone, ligament, tendon) (56, 104, 148, 544). Moreover, as athletes become faster, improving their physical capacity through resistance training should enable them to tolerate the higher joint loadings (39, 272, 301, 328, 410, 432, 544, 591), thus highlighting the inclusion of resistance training in an athlete's training programme.

Combined resistance training, and intersegmental control training during running and COD drills

King et al. (292) examined the effects of a rehabilitation programme which targeted intersegmental control in athletes with athletic groin pain. Athletes were subjected to three levels of rehabilitation: level 1 consisting of intersegmental control and strength training; level 2 focused on linear running drills focusing on lumbo-pelvic control and posture, and running mechanics; and level 3 focused on multidirectional technique drills which emphasised intersegmental control (holding a ball, or arms locked overhead) and lateral propulsion which was performed three times a week. Repeat 3D motion analysis revealed a 110˚ cutting task was performed with reductions in ipsilateral trunk side flexion (ES = 0.79), a factor linked to peak KAMs (127, 189, 267, 272), reduced hip abduction angle and hip adduction moment which has also been linked to greater peak KAMs (189, 245, 519), and increased pelvic rotation in the direction of travel (ES = 0.76) (Table 2.7). Furthermore, changes in variables connected with faster cutting performance were revealed including greater COM translation in the direction of travel relative to COP (ES = 0.40), reduced KFAs (ES = 0.33), and increased ankle plantar-flexor moments (ES = 0.48). While no differences in approach velocity were observed (p = 0.434, ES = 0.07), a slightly shorter GCT was noted (ES = 0.30), indicating potential performance benefits (143, 338, 346, 498). Unfortunately, KAMs or angles were not provided within the manuscript, though it is speculated the positive changes in lateral trunk flexion, hip abduction and hip adduction moment may indicate a reduction in peak KAMs (189, 245, 519). A note of caution is advocated, however, because there was no CG and measurement error values were not established.

Mixed programme – session performed separate from sports session that integrates exercises of at least three of the following modalities: trunk, balance, plyometric, strength training, flexibility

Yang et al. (622) recently examined the effects of a 4-week mixed training intervention programme consisting of trunk strengthening, stretching, balance training, hip extension strength training, and plyometrics in male and female basketball and volleyball players on 45° side-stepping. No statistically significant intervention effects on KFA, peak impact posterior GRF, or exit velocities were observed during a 45° cutting task (Table 2.7). As such, a 4-week mixed training intervention programme was ineffective in changing cutting biomechanics; however, 4 weeks could be a relatively short duration to potentially elicit positive adaptations, and it is worth noting that only three biomechanical variables were evaluated; thus, it is unknown what the effects were on frontal plane biomechanics, which are arguably of greater importance to injury risk (148, 183, 301). Moreover, a note of caution is warranted for the hip strengthening exercise repetitions prescribed by Yang et al. (622) because, although the authors describe the protocol as strength training, the repetitions/durations prescribed were in fact strength endurance (30 seconds of 1-2 sets). This is sub-optimal for eliciting maximum strength adaptations where low repetitions with higher loads would be required (215, 543).

Bencke et al. (33) compared the effects a 12-week prophylactic training programme on side-stepping GRF variables and muscle activity. The programme was performed twice a week, consisting of unilateral jump-landings, unilateral squats, hamstring pulls, hip abductions, and one-leg coordinated hopping in handball players in comparison to a CG who resumed normal skill training. Interestingly, the training intervention resulted in slightly greater VPFs (ES = 0.41), shorter GCTs (ES = 0.94) due to a shorter concentric phase duration (ES = 0.94), and a reduction in semi-tendinosis (ES = 0.63) and biceps femoris pre-activity duration (ES = 0.59) (Table 2.7). Therefore, the training programme had a positive effect on variables associated with faster COD speed performance such as greater VPFs (530, 531) and smaller GCTs (143, 338, 346, 498, 531), but the decreased hamstring muscle activity is of concern because high levels of hamstring muscle activation is needed to prevent anterior tibial translation and reduce anterior tibial shear (249, 322, 328, 341, 396, 496), thus ACL loading.

Weir et al. (594) demonstrated increases in total gluteal muscle activation and elevated contribution of hip extension moment to total support moment during unanticipated side-stepping following an 8-week mixed programme intervention (balance, plyometric, and body weight resistance training); however, no changes in frontal plane knee moments were observed in 13 female hockey players. Weir et al. (592) also demonstrated positive changes (reduced KIRM) in unanticipated side-stepping biomechanics following a 9-week high dosage mixed training intervention (balance, plyometric and resistance training) (4 × 20 min sessions), but no statistically significant changes in frontal plane moments for the whole group were observed. Recently, Staynor et al. (535) examined the effects of a mixed programme training intervention, based on the intervention by Weir et al. (592) (consisting of plyometric, resistance, and balance exercises, performed in-season twice a week for 9 weeks), on unanticipated side-stepping biomechanics in local female community-level athletes. KFMs increased post-training intervention (ES = 0.77), but no statistically significant changes in peak KAM and KIRMs were observed for the IG (ES ≤ 0.16), whereas the CG displayed greater KAMs and KIRMs (ES = 0.36-0.56) post-testing (Table 2.7). Additionally, the IG also produced kinematic changes associated with safer side-stepping cutting techniques such as reduced lateral foot plant distances, more erect trunk postures in the frontal plane, and increased knee flexion (ES = 0.40-0.84, Table 2.7). It is worth noting, however, that all mixed training programmes intervention studies have not acknowledged measurement error when interpreting their findings.

Consequently, based on the mixed training programmes intervention studies, it is inconclusive that this method of training is effective in reducing knee joint loading during COD. The results of these studies contrast to balance training (92, 93, 425) and COD technique modification interventions (126, 277) which have demonstrated reductions in COD knee joint loads. Although the mixed programmes did include balance exercises, the volume load and exercise duration of balances exercises were much lower than the successful interventions that solely focused on balance training. This discrepancy in volume load and duration may explain the contrasting findings. Additionally, it is speculated that the additional and combination of exercises from different modalities during these mixed programmes may interfere with balance training and may limit its effectiveness.

Dynamic core stability training

As the trunk contains over half of the body's mass, deficits in neuromuscular control and suboptimal trunk motion and position is a critical factor affecting knee joint loading (248, 366). Additionally, deficits in trunk control (i.e., core stability) have also been shown to be associated with non-contact ACL injury (635, 636). Consequently, several studies have investigated the effects of trunk conditioning on COD biomechanics (266, 602). Jamison et al. (266) reported what they defined as "combined resistance and trunk stabilisation" (which effectively involved solely static trunk exercises with resistance training) to be ineffective in reducing knee joint loads during cutting; however, in direct contrast, Whyte et al. (602) has recently demonstrated positive effects of a dynamic core stability intervention (i.e., trunk curls, dynamic bridges, planks, side planks, with added perturbations) on cutting mechanics (Table 2.6). Interestingly, following the 6-week intervention, athletes demonstrated increases in internal hip extensor moments and reductions in frontal and transverse knee joint loads (Table 2.6). This result is noteworthy because a combination of frontal and transverse knee joint loads can increase ACL loading to a greater extent than uniplanar loading (26, 513). Additionally, reductions in posterior GRF were observed as a result of the training intervention. This result is noteworthy because this adaptation may result in reductions in anterior tibial shear (506), thus injury risk (42, 43, 343, 615). Therefore, these findings indicate that dynamic core stability training could be an effective training modality to reduce ACL loading during cutting actions.

Surprisingly, trunk and pelvic kinematics remained unchanged following the intervention by Whyte et al. (602); thus, the successful reductions in knee joint loads could be partially attributed to the reduction in posterior GRF. While this a positive finding in terms of reducing potential ACL loading, the fact that posterior GRF reduced may negatively affect performance, because posterior GRF has been associated with faster COD performance (143, 202, 275). Unfortunately, Whyte et al. (602) did not examine cutting performance, but it is important to note that MLGRF will most likely be a larger contributing factor to faster cutting performance compared to posterior GRF (227, 228), but this was not examined in the study. Future research needs to consider both injury risk and performance implications to improve our understanding of the potential "performance-injury conflict" present during COD.

The successful results of dynamic core stability training are in direct contrast to Jamison et al. (266); however, these conflicting observations could be attributed to differences in exercise selection. For example, Jamison et al. (266) used static trunk stabilisation exercises, in contrast to dynamic core stability exercises used by Whyte et al. (602). The dynamic core stability exercises (with added perturbations) targets COM control and could be more specific to the trunk control requirements during cutting (599). It should be noted, however, that only one study has confirmed that dynamic core stability training is effective in reducing knee joint loading during COD. Further research is required to definitively confirm that this training method is effective in reducing COD knee joint loads.

Table 2.5. Summary of research that has investigated the effects of COD technique modification and COD speed and footwork training on COD biomechanics

Study	Subjects	Training intervention	COD task	Results (Post intervention)	Comments
COD technique modification drills					
Dempsey et al. (126)	Twelve male non-elite team sport (6 Australian football, 5 rugby union, and 1 soccer) athletes *3 withdrawals	6-week COD technique modification 2 x a week (15 min sessions) With immediate feedback (visual and oral)	45° ± 5° side-step. PP and UP ~5 m/s	At IC: • ↓ **lateral foot plant distance** ($p = 0.039$) PP (ES = 0.55), UP (ES = 0.58) • ↓ **lateral trunk flexion** ($p = 0.005$) PP (ES = 1.09), UP (ES = 0.14) At WA: • ↓ peak **KAM** ($p = 0.034$) PP (ES = 0.58) and UP (ES = 0.78) Both postural changes were correlated with the change in KAM • **Change in lateral foot plant distance** ($r = –0.468$, $p = 0.025$) • **Lateral trunk flexion** ($r = -0.377$, $p = 0.135$) ↔ in approach speed, knee flexion IC, and torso rotation	• No CG • Did not establish reliability, measurement error, or meaningful difference • Implications on performance unclear • Controlled approach velocity
Jones et al. (277)	Ten Female netball players	6-week COD technique modification 2 x a week Technique drills that encouraged PFC braking, backwards trunk inclination, and neutral foot position. Weeks: 1&2 – deceleration emphasis; 3&4 - Randomly with greater entry velocity; 5&6 - Drills performed randomly at speed unanticipated stimulus	180° turn - PP ~3 m/s	• ↓ **completion time** ($p < 0.05$, ES = 0.74) • ↓ **peak KAM** ($p < 0.001$, ES = 0.73) • ↓ **Initial foot progression angle** ($p < 0.001$, ES = 2.60) • ↓ **Initial trunk angle at FFC** ($p < 0.05$, ES = 0.58) • ↔ in approach velocity or horizontal GRF ratio (ES = 0.10-0.15) **Changes in initial foot progression angle and KAM** ($r^2 = 37\%$, $p = 0.028$)	• Athletes were not fast to begin with • No CG • Did not establish reliability, measurement error, or meaningful difference • Conference proceeding format
Change of direction speed and footwork					
Wilderman et al. (606)	30 female basketball players	6-week agility (COD speed, footwork, and manoeuvrability drills) – 4 x a week (N=15) And a CG (N=15)	45° side-step Sidesteps –PP 3.3-4.3 m/s	• ↑ **medial hamstring EMG activation** for IG (ES = 0.94) • ↔ in knee flexion angle and vertical GRF ($p > 0.05$, ES ≤ 0.15)	• Lack of feedback regarding COD technique • Absence of specific side-stepping drills

Key: ↑: Increase; ↓: decrease; ↔: no significant change; KAM: Knee abduction moment; IC: Initial contact; WA: Weight acceptance; IRM: Internal rotation moment; ROM: Range of motion; GCT: Ground contact time; BW: Body weight; GRF: Ground reaction force; PP: Pre-planned; UP: Unplanned; BW: Body weight; EMG: Electromyography; PFC: Penultimate foot contact; FFC: Final foot contact; ES = Effect size; CG: Control group; IG: Intervention group; COD: Change of direction; GRF: Ground reaction force;

Table 2.6. Summary of research that has investigated the effects of balance, dynamic core stability control training and perturbation-enhanced plyometric training on COD biomechanics

Study	Subjects	Training intervention	COD task	Results (Post intervention)	Comments
Balance training					
Oliveira et al. (425)	26 healthy men – recreational athletes	6-week Balance training – 4 x a week (30 mins) (n = 13) And a CG (n = 13)	90° cut and 1 unexpected perturbed cut (10cm translation) ~2.5 m/s	Balance group during perturbed cutting: • ↓ **peak KAMs** (33 ± 25%, $p < 0.03$, $\eta^2 = 0.487$) • ↑ **activation of trunk and proximal hip muscles** • ↑ **burst duration** prior (23 ± 11%) to landing ($p < 0.02$, $\eta^2 = 0.798$) • ↔ changes in peak force, approach and exit velocity ($p < 0.05$)	• Presents findings for the perturbed trial only, and this was for only 1 trial • Low approach velocity
Cochrane et al. (93)	Fifty male AFL players	Allocated either to a CG or to one of four 12-wk training programs. 1. Machine weights 2. Free weights 3. Balance 4. Machine weights and balances	30° and 60° side-step, 30° XOC PP and UP and– light delay ~4-4.5 m/s Preferred leg	Balance group • ↑ **Flexor/extensor contraction ratio** -18% • ↑ **Flexor muscle activation** • ↑ **Biceps femoris/semimembranosus co-contraction ratio** • ↓ **Quadricep activation** Strength training • ↓ **Flexor/extensor contraction ratio** and ↑ **Quadricep activation**	• Implications on performance unclear • Controlled approach velocity
Cochrane et al. (92)	Fifty male AFL players	Allocated either to a CG or to one of four 12-wk training programs. 1. Machine weights 2. Free weights 3. Balance 4. Machine weights and balance	30° and 60° side-step, 30° XOC PP and UP and– light delay ~4-4.5 m/s Preferred leg	Change in moments across WA in all manoeuvres (Mean and SD not provided, thus ES cannot be calculated): **Balance** • ↓ **peak KAM** ($p < 0.001$, 62%) and ↓ **peak IRM** ($p < 0.001$, 32%) in all manoeuvres **Free weights** • ↔ peak KAM and IRM **Machine Weights** • ↓ **peak KAM** ($p < 0.05$, 27%) **Machine weights + balance training** • ↔ peak KAM and IRM **CG** • ↑ **peak KAM** ($p < 0.05$, 26%)	• Did not establish reliability, measurement error or meaningful difference • Implications on performance unclear • Controlled approach velocity

Dynamic core stability training					
Whyte et al. (602)	31 male varsity footballers	6-week dynamic trunk control/core stability programme – 3 x a week (n = 15) And a CG (n = 16)	45° side-step PP and UP	**IG** • ↑ **internal hip extensor moment** (p = 0.017, η^2 = 0.079, 24-28% of stance)) for PP • ↓ **internal knee varus moment** (p = 0.026, η^2 = 0.076, 18%- 25% of stance) for PP • ↓ **knee external rotator moment** (p = 0.041, η^2 =0.066, 15%- 20% of stance) for PP • ↓ **posterior GRF** for both cuts ($p \leq 0.030$, η^2 =0.074-0.081) for PP and UP (11-30% and 15-19% of stance, respectively) • ↔ **in trunk and pelvic kinematics** (descriptive data not provided; thus, ES cannot be calculated)	• Use of SPM • Contains CG
Perturbation-enhanced plyometric training					
Weltin et al. (599)	28 females (soccer, handball, and basketball)* 4 withdrawals:	Perturbation-enhanced plyometric training (PPT) (n=12): lateral reactive jumps – 4-week training – 3 times a week Plyometric only – CG (n = 12)	45° side-step UP - 4.0±0.2 m/s	**PPT** • ↓ **trunk rotation** 7.2° (ES = 1.14), ↓ **step width** (p = 0.003, ES = 0.88), and ↑ **pelvic rotation** 4.1° (ES = 0.45) • ↓ **KAM** 0.05 Nm/kg, CG ↑ 0.14 Nm/kg (SD not provided; thus, ES cannot be calculated) • ↔ lateral trunk lean (ES = 0.26)	• Perturbation-enhanced method is unfeasible to implement in real world as it required motored platform

Key: ↑: Increase; ↓: decrease; ↔: no significant change; GRF: Ground reaction force; PP: Pre-planned; UP: Unplanned; ES = Effect size; CG: Control group; IG: Intervention group; SPM: Statistical parametric mapping; PPT: Perturbation and plyometric training; KAM: Knee abduction moment; IRM: Internal rotation moment; XOC: Crossover cut

Table 2.7. Summary of research that has investigated the effects of mixed training programmes and combined programmes on COD biomechanics

Study	Subjects	Training intervention	COD task	Results (Post intervention)	Comments
Mixed programme – session that integrates exercises of at least three of the following modalities: trunk, balance, plyometric, strength training, flexibility					
Weir et al. (592)	10 Elite female hockey players	16-week maintenance training program (3 x 10-minute sessions a week) which directly followed a high-dose 9-week initial training intervention (4 x 20 min sessions a week), as part of a biomechanically informed ACL injury prevention program. BW plyometric, resistance, and balance exercises	45° side-step - UP	• ↓ **peak KAMs** (ES = 0.30, 26.3%) following maintenance • High-risk responders displayed ↓ **peak KAM** (28.6%) and **IRM** (37.1%)	• Highlight the importance of continuing the training • Acknowledge there will be responders and non-responder • Abstract format
Weir et al. (594)	13 Elite female hockey players	8-week 4 x 15 min multicomponent sessions consisting of BW plyometric, resistance, and balance exercises	45° side-step - UP	• ↑ TMA of the gluteal (grouped maximus and medius) by 10% during WA ($p = 0.006$, power = 0.864). • $\boxed{\leftrightarrow}$ in frontal plane knee moments ($p < 0.01$, ES = 0.73), ↑ hip extension moment (ES = 0.56)	• No differences in frontal plane knee moments • Abstract format • No CG
Yang et al. (622)	22 male, 18 female (basketball and volleyball)	4-week multicomponent programme consisting trunk strengthening, stretching, proprioceptive training, hip extension strength training and plyometric training intervention - 3 x a week (N = 18, 9 male and 9 female) And a CG (N = 18, 9 male and 9 female)	45° side-step Sidesteps –PP – 5 step run-up	• $\boxed{\leftrightarrow}$ no differences in knee flexion angles, peak impact posterior GRF, or exit velocities compared to CG following intervention (4-12 weeks post training intervention)	• Multicomponent programme; however, strength exercises were prescribed for, strength/ muscular endurance • Considered only limited number of variables – unknown the effect of frontal plane biomechanics • Short duration
Bencke et al. (33)	17 male handball players	Mixed programme consisting of jump landings, unilateral squats, hamstring pulls, hip abductions, and one-leg coordinated hopping IG (n = 10) 12-weeks twice a week And a CG (n = 7)	Side-step (No other decsriptions provided)	IG • ↑ **VPF** (ES = 0.41), ↓ **GCTs** ($p < 0.05$, ES = 0.94) due to a ↓ concentric phase duration ($p < 0.05$, ES = 0.94) • ↓ **ST** ($p < 0.05$, ES = 0.63) and **BF** pre-activity duration ($p = 0.08$, ES = 0.59)	• No joint kinetics/kinematics examined

Staynor et al. (535)	25 Female community-level team sport athletes *6 withdrawals on training group	Split into IG (n = 8) and CG (n = 10), 2 x a week for 9 weeks (15-20 min sessions) Combination of BW plyometric, resistance, and balance exercises	Side-step – UP (full details not provided)	• IG ↑ **peak KFM** (ES = 0.77), $\boxed{\leftrightarrow}$ in **peak KAM** (ES = 0.16) and **IRM** (ES = 0.0), but **CG** ↑ **peak KAM** (ES = 0.36, 28%) and ↑ **IRM** (ES = 0.56, 38%) IG • ↓ **hip abduction** (ES = 0.70, 31%) • ↑ **knee flexion at foot strike** (ES = 0.59, 33%) • ↓ **trunk flexion range of motion** (ES = 0.97, 29%) • ↓ **lateral trunk flexion** (ES = 0.40, 16%) • ↓ **lateral foot plant distance** (ES = 0.84, 11%)	• Did not establish reliability, measurement error, or meaningful difference • Attendance and compliance rates of 71 ± 14 and 77 ± 7%
Combined balance and COD technique training					
Donnelly et al. (140)	AFL male athletes (n = 1001) 34 athletes for biomechanical testing (BTT, n = 20; ST, n = 14)	Balance and COD technique training (BTT) or acceleration training (ST) 2 x week – 20 mins week – 18 weeks 1 x week – weeks 19-28	45° ± 5°, side-step - PP and UP	Both training groups: • ↓ **peak IRM** (p = 0.025, ES = 0.57) -45% reduction, during PP • ↑ **peak KAM** (p = 0.022, ES = 0.44) - 31% increase during UP	• High athlete to coach ratio (40:1) • Low athlete compliance (45 ± 22%)
Combined trunk stabilisation and resistance training					
Jamison et al. (266)	22 males (previously played American football) N of 10 and 11 completed testing	RT only or Resistance and Trunk stabilisation 6 weeks - 3 sessions a week	45° ± 5°, side-step 3 steps self-selected jog	**RT only** • ↑ **peak KAMs** (p = 0.012, 50%) and ↑ **peak IRM** (p = 0.617, 12%) **Combined training** • ↑ **peak KAM** (p = 0.116, 35%) and ↓ **peak IRM** (p = 0.110, 35%) (SD not provided, thus ES cannot be calculated)	• Did not achieve *a priori* minimum sample size recommendations • Did not establish reliability, measurement error, or meaningful difference • Static trunk exercises were used
Combined resistance training, and intersegmental control training during running and COD drills					
King et al. (292)	112 athletes with athletic groin pain were assessed pre and post rehabilitation	Athletes were subjected to three levels of rehabilitation: • Level 1 intersegmental control and strength training • Level 2 on linear running drills (lumbo-pelvic control and posture) • Level 3 multidirectional technique drills which emphasised segmental control (using holding a ball, or arms locked overhead) and lateral propulsion	110° cut -PP, ~2 m/s	• ↓ **ipsilateral trunk side flexion** (ES = 0.79) • ↓ **hip abduction angle and hip adduction moment** • ↑ **pelvic rotation** in the direction of travel (ES = 0.76) • ↑ **centre of mass translation** in the direction of travel relative to centre of pressure (ES = 0.40) • ↓ **knee flexion angle** (ES = 0.33) • ↑ **ankle plantar-flexor moment** (ES = 0.48) • $\boxed{\leftrightarrow}$ in approach velocity (p = 0.434, ES = 0.07) • ↓ **GCT** (ES = 0.30) • ↑ **dorsi-flexion** (ES = 0.58) Large increase in total work done at the ankle, a moderate reduction in the total work done at the hip, and a small reduction at the knee after rehabilitation.	• Considered performance implications • Showed positive effects for injury risk and performance • No CG • Did not establish reliability, measurement error, or meaningful difference

Key: ↑: Increase; ↓: decrease; $\boxed{\leftrightarrow}$: no significant change; XOC: Crossover cut; KAM: Knee abduction moment; WA: Weight acceptance; IRM: Internal rotation moment; SD: Standard deviation; BW: Body weight; GRF: Ground reaction force;; PP: Pre-planned; UP: Unplanned; BW: Body weight; ACL: Anterior cruciate ligament; RT: Resistance training; ES = Effect size; CG: Control group; TMA: Total muscle activation; COD: Change of direction; VL: Vastus lateralis; BF: Biceps femoris; ST: Semitendinosus; VPF: Vertical propulsive force

Table 2.8. Summary of research that has investigated the effects of injury prevention warm-up protocols on COD biomechanics

Study	Subjects	Training intervention	COD task	Results (Post intervention)	Comments
F-MARC 11+ soccer specific warm-up					
Thompson et al. (556)	51 females aged 10-12 years soccer players *5 withdrawals	F-MARC 11+ (n = 26) 2 x a week for 7-8 week – 15 sessions total And CG (n = 20)	45° ± 5° Sidesteps -PP and UP ~4 m/s	**Bilateral jump (IG)** • ↓ **peak KAM** (p = 0.045, ES = 2.15) **Side-stepping (IG)** • ↑ **peak KAM** PP (p = 0.280, ES = 1.20, 10%) and UP (p = 0.044, ES = 1.98, 18%)	• Did not establish reliability, measurement error, or meaningful difference • Athlete compliance 70.2 ± 14.0%
Thompson-Kolesar et al. (555)	51 preadolescent females (28 intervention, 23 CG)* 5 withdrawals and 43 adolescent (22 intervention, 21 CG)* 6 withdrawals	F MARC 11+ (n = 26) 2 x a week for 7-8 week – 15 sessions total and CG (n = 20)	45° ± 5° Sidesteps -PP and UP ~4 m/s	**Preadolescents – PP side-step** • ↑ **precontact flexor-extensor muscle contraction** (p = 0.004-0.002) **Both groups – side-step** • ↔ in **knee valgus angles** or peak **KAM** - Inspection of graphs indicate ↑ peak KAMs in both groups (Descriptive data not provided so ES cannot be calculated)	• Highlights ineffectiveness of intervention for addressing cutting mechanics – only effective for bilateral task • Lack of volume and exercises that addresses COD mechanics with feedback, and lack of dynamic trunk exercises
Oslo neuromuscular injury-prevention warm-up					
Zebis et al. (639)	Elite handball (n = 8) and elite soccer (n = 12)	Oslo NMS warm-up intervention - 20 min warm up – one season	Side-step (No other decsriptions provided)	• ↑ **Pre-landing EMG activity ST** (p < 0.001, ES = 0.70-0.78) and activity at foot strike (p < 0.05, ES = 0.60) • ↔ **Quadriceps EMG** (ES = 0.10-0.23) • ↔ **Knee and hip joint angles** (ES = 0.11)	• Low sample size • Investigated low number of biomechanical variables • No CG
Zebis et al. (638)	40 adolescent female football and handball	12-week Oslo NMS warm-up – 3 x a week (n = 20) And a CG (n = 20)	Side-step (No other decsriptions provided)	IG • ↓ **VL-ST activity difference** (43%, p <0.0001) • ↑ **hamstring MVC** (p = 0.0134) • ↓ **VL EMG preactivity** (23%, p < 0.0008), ↑ **ST EMG preactivity** (18%, p < 0.0001), and ↑ **BF EMG preactivity** vs. CG • ↔ **peak KAM** or **knee valgus angle at IC** (Descriptive data not provided so ES cannot be calculated)	• Only frontal plane knee kinetics and kinematics

Core-Pac warm-up

Celebrini et al. (76)	Ten adolescent female soccer	baseline testing – acute changes (n=10) (move from the centre- lead with the belly button) 4 week – Core-Pac training intervention (n=7) 4 x week	15-55° Side-step PP and UP	5 of 7 subjects displayed **↑ knee flexion angle and ↓ peak KAM**	• Individual differences in response to training intervention • No CG
Celebrini et al. (75)	Twenty adolescent female soccer	6 week – Core-Pac training intervention – 4 x a week (n=10) And a CG (n=9)	15-55° Side-step PP and UP	IG • ↑ **knee flexion angle** PP cutting (*p* = 0.001, ES = 2.02) • $\boxed{\leftrightarrow}$ **peak KAM** for PP and UP (Descriptive data not provided so ES cannot be calculated)	• Low sample size • No immediate feedback regarding their technique or biofeedback

Key: ↑: Increase; ↓: decrease; $\boxed{\leftrightarrow}$: no significant change; KAM: Knee abduction moment; IC: Initial contact; IRM: Internal rotation moment; GCT: Ground contact time; BW: Body weight; NMS: Neuromuscular; PP: Pre-planned; UP: Unplanned; EMG: Electromyography; RT: Resistance training; ES = Effect size; CG: Control group; IG: Intervention group; COD: Change of direction; SD: Standard deviation; VL: Vastus lateralis; BF: Biceps femoris; ST: Semitendinosus; VPF: Vertical propulsive force; MVC: Maximal voluntary contraction; Core-Pac: Core position and control; F MARC 11+: FIFA NMS warm-up; IC: Initial contact

Perturbation-enhanced plyometric training

Weltin et al. (599) investigated the effects of perturbation-enhanced plyometric training (lateral reactive jumps on a motorised platform which moved), in comparison to regular plyometric training in female athletes. Interestingly, four weeks post-intervention, the perturbation-enhanced plyometric group athletes displayed reductions in trunk rotation and decreases in step width (Table 2.6), both of which are associated with greater KAMs (148, 183, 189, 301). Although not statistically different, the perturbation-enhanced plyometric group showed a slight reduction in KAMs, while KAMs increased in the plyometric training only group (Table 2.6). Surprisingly, lateral trunk lean remain unchanged following the perturbation-enhanced training; however, this absence could be attributed to the lack of feedback and cueing regarding trunk control in contrast to previous studies that have found positive changes in lateral trunk lean (126). Consequently, perturbation-enhanced lateral reactive jump training reduces characteristics (trunk rotation and step width) associated with greater peak KAMs during directional changes but appears to be ineffective in producing statistically significant reductions in peak KAMs. Thus, more research is required around plyometric related interventions for the development of safer cutting mechanics.

Injury prevention warm-up training protocols

Given the simplicity of training exercises to be integrated into the warm-ups of field-based sessions for athletes to improve neuromuscular control, and its relative success in reducing ACL injury rates (241, 398, 432, 589), several studies have investigated the effects of the Oslo, Core-Pac, and F-MARC 11+ warm-up training interventions on COD biomechanics (Table 2.8). These interventions involve a 15-25 minute protocol that is performed prior to sport-specific practice (i.e., technical and tactical) and / or games.

Oslo neuromuscular warm-up intervention

Zebis et al. (639) found the Oslo warm-up training intervention increased pre-landing semitendinosus activity ($p < 0.001$, ES = 0.70-78), but unchanged quadriceps activity, hip and knee joint angles (ES = 0.10-0.23) during a side-stepping task in female handball and soccer players (Table 2.8). It is worth noting, however, that there was no CG, and only a limited number of biomechanical variables were examined (hip and knee joint angles, EMG activity). Including a CG, more recently, Zebis et al. (638) examined the effects of the Oslo neuromuscular warm-up protocol on side-stepping biomechanics and EMG muscle activity. The IG displayed a potentially safer agonist-antagonist muscle pre-activity pattern, with elevated semitendinosus and biceps femoris pre-activity, and a reduction in vastus lateralis activity post-training, in contrast to the CG (Table 2.8). This finding is noteworthy because a lack of pre-activity observed with the medial hamstrings in combination with greater proportion of lateral quadriceps recruitment may compress the lateral joint, open the medial joint, increase knee valgus, increase anterior shear force, and therefore increase ACL loading (249, 322, 341, 396, 496). For instance, in a cohort study, athletes who went on to injure their ACL displayed higher vastus lateralis pre-activity and reduced semitendinosus activity compared to uninjured athletes during a cut (637). It is worth noting that Zebis et al. (638) observed no changes in peak KAM or knee valgus angles at IC following the training intervention, but unfortunately the authors failed to present the mean and standard deviations, thus the effect size could not be established. Consequently, the Oslo-warm-up protocol produces favourable agonist-antagonist muscle pre-activity patterns but appears to have a negligible effect on frontal plane knee moments.

Core-Pac warm-up intervention

In light of the positive effects regarding the within-session changes in cutting technique adopting Core-Pac movement strategy (76), in the same study the authors also investigated the chronic effects of Core-Pac warm-up training intervention in female soccer players. The warm-up consisted of balance, trunk, lower-limb control, multidirectional running, and COD drills. Due to a low sample size, statistical analysis was not performed; however, five of seven subjects displayed increases in KFA and reduced peak KAMs during cutting following the training intervention (Table 2.8). It is worth noting, however, that there was no CG, but the preliminary results highlight the individual variation in response to training interventions. Expanding on their previous work, Celebrini et al. (75) compared the effects of the Core-Pac warm-up in comparison to a CG who completed a normal warm-up routine. Following a six-week intervention, the female soccer players who participated in the Core-Pac displayed an increased KFA during cutting (p = 0.001, ES = 2.02) (Table 2.8), but peak KAM remained unchanged. The researchers stressed two notes of caution: firstly, the study contained a low sample size and may therefore lack statistical power; and secondly, there was an absence of coaches' and technological feedback regarding technique which may explain the ineffectiveness in reducing peak KAM. The absence of feedback is in contrast to previous studies which have provided immediate feedback and subsequent successful reductions in knee joint loading (126, 277). Consequently, further research is needed to confirm the efficacy of the Core-Pac training intervention on COD knee joint loading.

F-MARC 11+ soccer specific warm-up

Thompson et al. (556) investigated the effects of the F-MARC 11+ soccer specific warm-up on biomechanical risk factors associated with ACL loading in preadolescent female soccer players. The soccer players were divided into a control and IG, with the neuromuscular warm-up performed two times a week for seven to eight weeks. Of concern, moderate to large increases ($p \leq 0.044$, ES = 1.18-1.95), in peak KAMs were demonstrated during pre-planned and unanticipated cutting (Table 2.8). Unfortunately, cutting performance was not examined, thus the implications of the F-MARC 11+ training intervention on performance is unclear. Critically, the F-MARC 11+ intervention was ineffective in reducing peak KAMs during side-step cutting. This finding is noteworthy because cutting actions are associated with non-contact ACL injury, particularly in soccer (94, 298, 310, 376, 583).

More recently, Thompson-Kolesar et al. (2018) has also confirmed that the F-MARC 11+ soccer specific warm-up was ineffective in reducing peak KAMs or knee valgus angles during cutting tasks in adolescent athletes (Table 2.8), substantiating the results of their earlier study in preadolescents. This observation could be attributed to the lack of repetitions and volume of COD technique training in the programme. The F-MARC 11+ programme primarily consists of bilateral tasks such as squats and jump-landings that are integrated with balance and trunk conditioning, which could explain why KAMs only reduced during the bilateral drop landing task only. Conversely, the technique modification intervention by Dempsey et al. (126) involved 15-minutes exclusively of COD technique modification, thus greater specificity and volume, resulting in reductions in KAMs. Therefore, these findings suggest the F-MARC 11+ does not adequately address deficits in cutting biomechanics in preadolescent and adolescent athletes but appears to be effective in reducing knee joint loading during bilateral landing activities. It is worth highlighting, however, that overall the F MARC 11+ has been shown to reduce ACL injury rates in male collegiate soccer players when implemented correctly (521), potentially due to positive adaptations in strength and muscle activation which may assist in ACL injury reduction. These adaptations have been observed following neuromuscular warm-up interventions (241, 432). Nevertheless, in terms of reducing knee joint loads during COD, the FIFA 11+ is an ineffective training modality, but globally may warrant inclusion into an athlete's holistic training programme to assist in ACL injury mitigation.

<u>*Maintenance training*</u>

While reductions in biomechanical characteristics associated with ACL injury risk have been demonstrated with various training modalities (93, 115, 126, 127, 277, 592), it is also important to understand the training dosages required to retain the improved movement biomechanics and reduced knee joint loads following the training intervention. To the best of the author's knowledge, only one study has examined the effects of performing dosages of maintenance training following a period of high dosage mixed training. Weir et al. (592) demonstrated positive changes (reduced KIRM) in unanticipated side-stepping biomechanics following a 9-week high dosage multicomponent training intervention (balance, plyometric and resistance training) (4 × 20 min sessions), and found a 16-week maintenance training programme (3 × 10 minute sessions) resulted in meaningful reductions in peak KAM (-26.3%, g = 0.30) (Table 2.7). As expected, the maintenance programme was particularly effective in retaining improved side-stepping biomechanics in the responder/"high-risk" group (classified as moderate-large effect size change) (Table 2.7). As stated previously, only one study has examined the effects of maintenance training dosages of on COD biomechanics; thus, making it difficult to establish maintenance training guidelines. Consequently, more longitudinal studies are required which investigate the effects of maintenance training on COD biomechanics to improve our understanding regarding the maintenance of improvements in COD biomechanics.

2.3.5.4 CONCLUSION

Based on the literature (Tables 2.5-2.8), balance training (92, 93, 425) is a potentially effective strategy to reduce knee joint loads during cutting; most likely attributed to eliciting safer knee agonist-antagonist muscle patterns and hip and trunk muscle activity. These positive biomechanical and neuromuscular adaptations may partially explain why balance training has been shown to reduce ACL injury rates (72, 398). COD technique modification (76, 115, 126, 127, 277) also appears to be an effective training strategy for addressing COD biomechanical deficits associated with increased ACL loading and therefore potential non-contact ACL injury risk. It should be noted, however, that the COD technique modification interventions which have shown promising results have not contained a CG and, as such, is a recommended area of further research. Moreover, the effectiveness of COD technique modification training on ACL injury rates has yet to be investigated. Nevertheless, in order to reduce knee joint moments and subsequent ACL loading, the magnitude of the GRF or moment arm must reduce (301). As such, practitioners interested in reducing COD knee joint loading for their multidirectional athletes should consider incorporating balance and COD technique modification into their athletes' training programmes to reduce potentially hazardous knee joint loads when changing direction.

One study has shown promising results regarding the effectiveness of dynamic core stability training on COD knee joint loading (602), but further research is needed to definitively confirm the efficacy of this method. Perturbation-enhanced plyometric training (599), the F-MARC 11+ (555, 556), Oslo Neuromuscular warm-up protocol (638, 639) and resistance training (93, 266) are ineffective in reducing COD knee joint loads, whereas conflicting findings have been observed for the Core-Pac (75, 76), and mixed programme training interventions (33, 292, 535, 592, 594, 622). More research is required around plyometric related interventions for the development of safer cutting mechanics. Although several studies have shown mixed training programmes and neuromuscular training appear to be ineffective in addressing COD biomechanics associated with increased ACL loading and potential non-contact injury risk (Tables 2.7 & 2.8), these training modalities have been shown to be effective in reducing ACL injury rates (241, 398, 432, 589) and may improve other qualities such as strength, muscle activation, and athletic performance (241, 432). Similarly, resistance training appears to be ineffective for reducing COD knee joint loads; however, this training modality elicits positive performance adaptations (284, 543, 544, 587) and is considered important for athletes to tolerate the loading associated when changing direction (39, 328, 410, 432, 544, 591). Therefore, mixed training programmes, injury prevention neuromuscular warm-ups, and resistance training should not be overlooked, and warrant inclusion into an athlete's holistic training programme.

Finally, to understand the most efficacious training modalities for addressing COD biomechanics associated with increased ACL loading, further research is needed in larger samples sizes, while containing a CG, and acknowledging measurement error to establish "real" and "meaningful" changes. Given the potential "performance-injury conflict" during COD (148, 183, 228), future studies need to consider the implications of the training intervention on both performance (completion time, GCT, exit velocity) and injury risk biomechanics to better inform injury risk mitigation programmes, because athletes may be unlikely to adhere to training programmes which negatively affect performance.

2.3.6 Additional factors that affect COD biomechanics (angle, velocity, limb dominance, anticipation)

A narrative review discussing the effect of angle and velocity on COD biomechanics has been published (148) (Appendix 8.1), thus a brief overview is provided here. It is important to note that biomechanical demands of COD can be described as "angle dependent" (39, 108, 212, 213, 227-229, 362, 411, 504, 505, 516) and "velocity dependent" (114, 290, 301, 403, 574) and are both critical factors that affect the technical execution of directional changes, deceleration and reacceleration requirements, knee joint loading, and lower-limb muscle activity. Thus, these two factors regulate the progression and regression in COD intensity. Specifically, faster and sharper CODs elevate the relative risk of injury due to the greater associative knee joint loading (39, 108, 228, 290, 301, 403, 505, 516, 574); however, faster and sharper CODs are key manoeuvres for successful performance in multidirectional sport, which subsequently creates a "performance-injury conflict" for practitioners and athletes. Furthermore, an "angle-velocity trade-off" exists during COD, whereby faster approaches compromise the execution of the intended COD (504, 548, 574).

From both performance and risk of injury perspectives, it would be advantageous for athletes to have the capacity to change direction safely and quickly from both limbs, given the unpredictable nature of multidirectional sports. A systematic review has been published regarding the effect of limb dominance on COD biomechanics (145) (Appendix 8.1), thus a brief overview is provided here. Generally, limb dominance is defined as the limb an athlete would prefer to kick / throw a ball with, but this approach is flawed because it is difficult for some athletes to delineate a preferred kicking / throwing limb, particularly if they are unfamiliar with a kicking or throwing action. However, more importantly, it has been reported that limb dominance as indicated by a preferred kicking limb does not necessarily equate to faster COD performance from that push-off limb (150). Thus, it has been recommended that limb dominance in the context of COD is defined as the limb an athlete would prefer to push-off / COD from (145). Nevertheless, the main findings from the systematic review (145) were that female soccer players, male rugby players, and female handball players exhibit subtle side-to-side differences when performing cutting manoeuvres (32, 66, 207, 345, 370, 456). The limb displaying "high-risk" mechanics is inconsistent within- and between studies and populations, thus it is inconclusive whether a particular limb is of heightened injury risk during side-step cutting.

Two systematic reviews (9, 65) have confirmed that compared to pre-planned directional changes, unanticipated directional changes evoke "higher-risk" knee mechanics associated with increased knee joint loading, thus risk of ACL injury. Specifically, greater sagittal (38, 126, 289), transverse (38, 126), and frontal (38, 98, 140, 289, 318, 381) plane knee joint loading have been demonstrated during unanticipated directional changes, while Weinhandl et al. (590), via musculoskeletal modelling, reported significantly greater ACL loading during unanticipated side-steps through combined increased sagittal (62%), frontal (26%) and transverse (12%) plane loading. Moreover, greater lateral trunk flexion during unanticipated directional changes has been observed compared to pre-planned (318, 319, 381); a high-risk posture associated with greater knee joint loads (126, 127, 189, 267, 272, 593).

It has been observed that non-contact ACL injuries occur when an athlete has externally focused attention in response to a stimulus with high visual and spatial-temporal constraints, such as evading or reacting to an opponent or reacting to a ball (54, 63, 204, 376, 426, 541). The deleterious effects of unanticipated directional changes through the introduction and response to stimuli increases the task complexity and imposes temporal constraints on the central nervous system (38, 98, 381), which reduces the preparation time for athletes to appropriately adjust their whole-body posture (318, 319, 381). This in turn, may explain the propensity to generate greater multiplanar knee joint loads during unanticipated COD which can potentially increase ACL strain (288, 342, 424, 512, 513). It is worth acknowledging that the majority of the unanticipated COD studies have used flashing arrows/ lights as an external stimulus; however, the timing and type of stimuli can affect COD biomechanics and subsequent knee joint loading (192, 318, 319, 381). Additionally, the use of flashing lights/arrows as an unanticipated stimulus have been criticised because they are not a sport-specific stimulus (411, 446, 626). Athletes do not react to flashing lights and arrows in sport, instead they scan and process visual and kinematic cues regarding the environment, sport, and athletes when performing CODs (268, 626). Researchers have shown reactive agility drills using flashing lights/arrows does not differentiate skilful performers (626-628), and in fact is a more complex and hazardous task compared to reacting to 2D video footage (318).

Unfortunately, it is difficult to provide a controlled (timing), 3D sport-specific stimulus in the laboratory when assessing unanticipated COD biomechanics. This has led to researchers using sport-specific COD tasks with externally directed attention during a side-stepping task such as attending to a ball (79, 174), side-stepping past an opponent (359), or carrying an object (ball or lacrosse stick) (84) which can evoke "higher-risk" postures and increase knee joint loading compared to pre-planned tasks. These findings may partially explain why the ball carrier in rugby (376) and handball (426) is commonly injured when side-stepping past an opponent, and the ACL injured limb is predominantly the racket hand side during plant-and-cut actions in badminton (291).

2.4 Methodological and technical issues in three-dimensional motion and ground reaction force analysis

3D motion and GRF analysis is considered the gold standard for evaluating movement kinetics and kinematics (187, 241). A diverse range of biomechanical parameters can be examined in order to evaluate an athlete's mechanics in relation to injury risk and performance, with various data collection and analysis methodologies available to biomechanists (208, 447, 486). It is imperative, however, that researchers and biomechanists are aware of the methodological and technical factors that can affect the outcome measure values, reliability, and subsequent evaluations of an athlete's biomechanical profile (Figure 2.10). Figure 2.10 outlines methodological and technical issues in evaluating COD biomechanics and this review will primarily focus on low-pass filtering COF, trial size, and data analysis procedures because these issues form the methodological studies of the thesis.

Figure 2.10. Methodological and technical issues in evaluating COD biomechanics (COF: Cut-off frequencies; COD: Change of direction)

2.4.1 Effect of low-pass filtering cut-off frequency on joint moments during dynamic activities

The analyses of joint moments at high speed are important for understanding human motion from both athletic performance and risk of injury perspectives. Joint moments correspond to resultant muscle forces and loading of passive structures, thus providing an insight into potential mechanisms of sporting related injuries. For example, KAMs, KIRMs, and KFMs are used as surrogates of ACL loading

(39, 126, 631). The combination of GRF and segment accelerations (marker data: the reflective markers attached to the participant), with estimates of mass and inertial properties, are used to permit calculations of joint moments through inverse dynamics (47, 305, 618). The marker (kinematic) and GRF (kinetic) data are typically filtered in an attempt to remove noise and preserve as much of the signal as possible; this can be achieved with different methods including polynomial smoothing, cubic and quintic splines, fourier smoothing, moving averages, and digital filtering (Butterworth) (486, 618).

Variation in low-pass filtering COF selection

Digital filtering is typically performed within numerous COD, running, DVJ, and landing investigations and commonly implemented is a fourth-order low-pass Butterworth filter for the smoothing of marker and force data (5, 66, 126, 127, 207, 227-229, 272-274, 523). However, a contentious issue in the computation of joint moments, during high velocity movements, is the selection of appropriate COFs for the smoothing of marker and force data. Currently, there is no consensus on the optimal combination of low-pass filtering COFs to attain accurate and reliable joint moments via inverse dynamics (44, 47, 304, 305, 351, 493). A broad range of matched and mismatched COF combinations have been reported across the COD, running, DVJ, and landing literature, while some studies only report COF for marker data, and a few studies fail to report any filtering procedures (Table 2.9).

Table 2.9. Cut-off frequency combinations used in the movement biomechanics literature

Matched COF		Mismatched COF			
COF	**Study**	**COF**	**Study**	**COF**	**Study**
10 Hz	(585)	7 and 15 Hz	(207)	10 and 200 Hz	(114)
12 Hz	(132, 179)	7 and 25 Hz	(108)	12 and 200 Hz	(227-229)
15 Hz	(101, 141, 301, 302, 346)	8 and 50 Hz	(66, 79, 455)		
18 Hz	(93, 125-127, 488)	9 and 50 Hz	(242, 416)		
20 Hz	(337, 497, 574, 590, 591)	12 and 25 Hz	(5, 272-274)		
30 Hz	(556, 561)	12 and 50 Hz	(98, 523)		
50 Hz	(551)	20 and 30 Hz	(280)		
Studies only reporting COF for marker data (259, 356, 358, 359, 362, 421, 516, 518, 564)					
Studies failed to state filtering procedures (84, 262, 360, 400)					

Key: COF: Cut-off frequency

Effect of low-pass filtering cut-off frequency on joint moments during dynamic activities: recommendations and considerations

Notably, the COF selection for marker and force data can impact the resultant lower-limb joint moment value obtained and may also influence the interpretation of an athlete’s biomechanical profile (44, 47, 304, 305, 493, 558). For example, Kristianslund et al. (304) reported significantly greater ($p < 0.0001$) knee flexor, hip flexor and abductor moments, and KAMs with mismatched COFs (10-50 and 15-50 Hz) compared to matched COFs (10-10 and 15-15 Hz) during a sport-specific cut. Likewise, Bezodis et al. (44) observed fluctuations in resultant knee joint moments during sprinting when filtering COFs were mismatched, similar to previous research findings that demonstrated greater knee moments with mismatched COFs (47). Roewer et al. (493) also demonstrated mismatched (10-50, 12-50 and 15-50 Hz) COFs produced significantly greater KAMs during DVJs compared to matched COFs (10-10, 12-12 and 15-15 Hz). Moreover, McCaw et al. (351) also found low-pass filtering to affect ankle, knee, and hip joint powers and moments during drop landings, while Tomescu et al. (558) has recently shown that low-pass filtering impacts muscle force estimates, contact forces, and joint moments during a single-leg jump-landing. As such, it has been recommended that the same matched combination of filtering COFs is used for both marker and force data to avoid artefacts and minimise oscillations in the moment curves (44, 47, 304, 305).

Contrary to the recommendations of previous research (44, 47, 304, 305), particularly the recommendations of Kristianslund et al. (304, 305), numerous biomechanical investigations have continued to implement mismatched COFs for marker and force data (5, 66, 98, 114, 207, 227-229, 272-274, 523). This absence could be explained by the findings and recommendations of Roewer et al. (493), who express a note of caution when applying matched COFs when smoothing marker and force data. They suggest that the impact peaks are real forces which must be absorbed through active and passive structures of the body, and smoothing out the impact peaks in an attempt to produce smoother curves may "obscure physiologically meaningful information, and lead to artificially small KAM values and costly false-negative findings." In addition, when interested in GRF variables, such as braking and propulsive force and impulse, a higher COF may be administered to avoid over smoothing and to preserve the signal to a greater degree (58, 162, 351, 558).

Effect of mismatched COF on ranking and classification of subjects

COF selection can also impact the ranking and classification of athletes in terms of KAM (304, 493). Kristianslund et al. (304) found 20 players were identified above a KAM threshold (mean + one standard deviation) during a 10-50 Hz condition, but only 11 players were still considered "high-risk" when applying the matched combination 10-10 Hz. Similarly, Roewer et al. (493) reported the rank order of KAMs attained from mismatched filtering COFs significantly covaried with the rank order attained from matched filtering COFs. Seventeen of the 22 subjects were identified as "at-risk" using the 10-50 Hz COF (one or more knee >25.25 Nm KAM threshold); however, three subjects were not identified as "at-risk" using the matched COF. Thus, it could be argued that applying the same COF for both marker and force data could potentially fail in the identification of "at-risk" athletes resulting in "false-negative" findings while alternatively, mismatched COF combinations could arguably produce "false-positive" findings.

Effect of mismatched COFs on KAM curve is inconsistent

It should be noted that the effect of mismatched COFs on the joint moment curve is not always consistent across participants, with individual variation present (493). As illustrated in Figure 2.11, a mismatched COF produced a greater spike and oscillation in the KAM curve for participant B, whereas an attenuated and blunted response is present for participant A. Thus, the statements made previously by researchers that mismatched COF combinations produce large spikes in joint moments (artefacts) should be interpreted with caution and the individual variability should be acknowledged when computing joint moments.

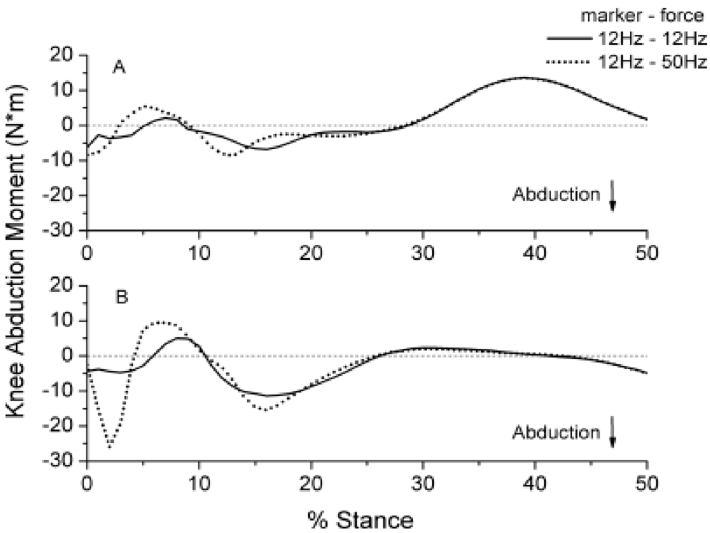


Figure 2.11. External KAM for two participants using the matched and mismatched COFs illustrating individual variation. Taken from Roewer et al. (493)

Taken together, the lack of consensus regarding the optimal COFs for the filtering of marker and force data creates a problem for biomechanists attempting to evaluate KAMs and joint moments during high velocity and impact tasks such as directional changes, sprinting, and landings. Further insight is required into the effect of different low-pass filtering COF combinations on high velocity and impact joint moments during dynamic activities.

2.4.2 Effect of trial size on biomechanical outcomes: implications on reliability and stability of mean

Biomechanists and researchers commonly evaluate a participant's movement mechanics using multiple trials, and then the subsequent key variables are then averaged. To ensure that the averaged values are valid, stable, and reliable representations of the participant's typical movement, it is imperative that the participant performs the optimal (minimum) number of trials (200, 220, 257, 264, 370, 374, 383, 553). Inspection of a single trial, or best trial, can lead to invalid data and erroneous conclusions (23, 264), and it has been reported that for sample sizes of 20, 10, and 5, trial sizes of 3, 5, and 10, respectively, provide sufficient statistical power (23). Due to the issues of movement variability (22, 466, 570), selecting too few trials may not represent the individual's long-term performance and therefore, it may be prudent to select a high trial size to provide stable and reliable data (264). However, it should be noted that selecting a trial size which is too high may increase the susceptibly of fatigue, reduced motivation, potential pacing strategies, and learning effects, while the associated time (additional time for data collection) and financial implications should also be acknowledged with increased trial sizes (89, 181, 200, 257, 383). As such, trial size is an important methodological consideration in the design of human movement experiments.

Unfortunately, most human movement biomechanical studies provide little rationale and justification for the trial size adopted (200, 264, 553). This finding is problematic because trial size can influence the stability, validity, and reliability of data (200, 220, 257, 264, 370, 374, 383, 553). The optimal trial size has been reported to be dependent on the task, variable, anatomical plane, and statistical method used (Table 2.10). Table 2.10 outlines the various trial size recommendations for a range of dynamic tasks including walking, running, jumping, landing, cricket bowling, soccer kicking, and hopping.

Table 2.10. Trial size recommendations for dynamic tasks

Study	Task and number of trials performed	Variables	Statistical approach	TS recommendations
Lees & Rahnama (320)	Instep soccer kick - 20 trials	Range of kinetics and kinematics	SD, CV%, SEM%	10-15
Amiri-Khorasani et al. (10)	Soccer kick - 10 trials	Kicking mechanical variables	ANOVA	5
Mok et al. (370)	COD - 5 trials	Range of lower-limb kinetics and kinematics	ICC	3
Monaghan et al. (374)	Walking gait - 10 trials	Range of lower-limb kinetics and kinematics	ICC, Bland and Altman, SD	10
Hamill & McNiven (220)	Walking gait - 25 trials	GRF	SET (0.25 SD bandwidth)	10
DeVita & Bates (129)	Walking gait - 50 trials	GRF	Comparison of mean, SDs and SET (0.10, 0.25, and 0.50 SD bandwidth)	25
Preatoni et al. (465)	Race walking - 40 trials	Range of kinetics and kinematics	CV%, SET (0.25 SD bandwidth)	15
Bates et al. (24)	Running - 10 trials	GRF	SET (0.25 SD bandwidth)	8
Diss (134)	Running - 10 trials	Range of lower-limb kinetics and kinematics	ANOVA Spearman-Brown prophecy	10
Rodano & Squadrone (491)	Vertical jumping - 25 trials	Lower-limb joint moments and powers	SET (0.30 SD bandwidth)	12
Racic et al. (471)	Jumping - 20 trials	GRF	ICC, SET (0.25 SD bandwidth)	4-9 (ICC), 10 (SET)
James et al. (264)	Landings - 20 trials	GRF	ICC, SET (0.25 SD bandwidth)	4-8 (ICC), 12 (SET)
Gore et al. (200)	Lateral hurdle hop -15 trials	Range of whole-body kinetics and kinematics	ICC, SET (0.25, 0.30, and SEM bandwidths)	6 (dependant on joint, statistical method and anatomical plane) - ICC generally lower recommendations
Chua et al. (89)	Basketball lay-up - 10 trials	Foot loading characteristics	ICC, SET (0.25 SD bandwidth)	8 (SET), ICC insensitive to inter-trial differences
Taylor et al. (553)	Overarm throwing - 30 trials	Upper body kinematics	SET (0.25 SD bandwidth), ANOVA	13-17
Stuelcken et al. (542)	Cricket bowling - 20 trials	GRF	SET (0.25 SD bandwidth)	12

Key: ICC = Intraclass correlation coefficients; SET= Sequential estimation technique; CV% = Coefficient of variation; SEM% = Standard error of mean; GRF = Ground reaction force; COD = Change of direction; TS = Trial size; ANOVA: Analysis of variance; SD: Standard deviation

Importantly, the statistical method used is a key factor when determining trial size recommendations (89, 200, 383, 471). Table 2.10 outlines the different statistical approaches adopted across the human movement literature. In most cases, recommendations based on ICCs result in lower trial size recommendations, while recommendations based on a sequential estimation technique (SET – alternatively known as sequential analysis technique) are higher and lead to more conservative recommendations (Table 2.10). The SET is used to determine minimum trial size based on an acceptable estimate of the stability of the mean (200, 220, 553, 581). This technique involves using a cumulative mean (moving point mean-adding one trial at a time) and a predefined bandwidth whereby stability is determined when the cumulative mean falls within the bandwidth (commonly - total trial mean ± 0.25 SD), and remains there for all subsequent trials (200, 220, 553, 581) (Figure 2.12). However, it is worth noting that the SET method may be limited by its use of an arbitrarily defined bandwidth, typically ± 0.25 SD, but ± 0.3 SD has also been used, and the number of reference trials can also influence the result (200, 264, 553). Gore et al. (200) proposed an alternative approach for SET analysis by using the standard error or measurement (SEM) to replace the SD bandwidth to establish true changes and differences in performance. It is worth noting, however, that the SEM is still influenced by the ICC, but it may provide a more conservative estimate compared to the traditional ± 0.25 SD bandwidth.

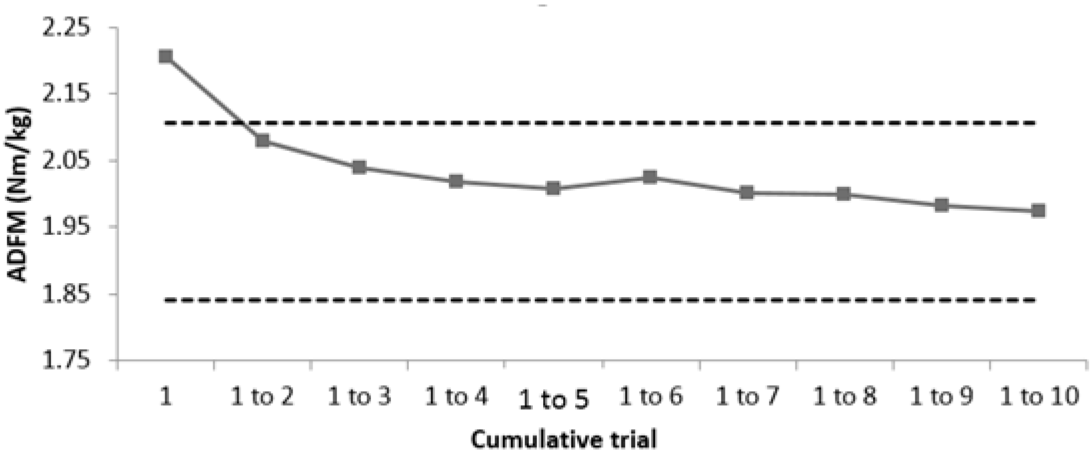


Figure 2.12. SET outlining stability in cumulative mean in reference to 10 trial SD bandwidth

2.4.3 Influence of trial size on change of direction biomechanics

A diverse range of trial sizes have been used to examine COD biomechanics (Table 2.11); however, the justification of these trial sizes is not reported, and the effect of trial size on outcome values and reliability of COD kinetics and kinematics remains relatively unclear. To the knowledge of the author, only one study has examined the effect of trial size on COD lower-limb kinetics and kinematics, observing only minor improvements in ICC values (ICC = 0.91 to 0.95 within-session and 0.73 to 0.77 between-session) when increasing trial size from three to five (370). However, the authors only compared ICC measures between trials sizes, failing to consider the effect of trial size on the outcome values, and failed to examine the stability of the mean using the SET technique as adopted in other trial size investigations (200, 220, 264). Therefore, it is uncertain whether there were any systematic differences (bias) between trials (i.e., potential learning, motivation, or fatiguing effects) (16, 27, 254, 335). Furthermore, Mok et al. (370) only considered increasing trial size from three to five, although some researchers have used 10 trials for evaluating COD biomechanics (38, 358, 359, 362) (Table 2.11), and this therefore warrants further investigation.

Table 2.11. Trial sizes used in the COD literature

Trial size	Study
Three	Kristianslund et al. (301), Vanrenterghem et al. (574), Jamison et al. (267), Chaudhari et al. (84)
Four	Havens and Sigward, (227, 228), Dempsey et al. (126)
Five	Dempsey et al. (127), Brown et al. (66), Dai et al. (114, 115)
Six	Jones et al. (272-274)
Ten	Besier et al. (39), McLean et al. (358, 359, 362)

Collectively, based on the aforementioned findings, trial size is an important factor that should be considered in methodological research designs due to its effect on the validity, stability, and reliability of human movement biomechanical data (200, 220, 264, 370, 374, 383, 553) (Table 2.10). Optimal trial size recommendations have been provided for various dynamic tasks such as walking, running, throwing, jumping, and kicking (Table 2.10); however, there is no clear consensus on the optimal trial size to obtain accurate and reliable COD biomechanical data. Biomechanists require information regarding the minimum trial size required to obtain a reliable and accurate assessment of an athlete's COD biomechanical profile. As such, further investigation exploring the effects of trial sizes from three to as high as 10 on outcome values, and the examination of reliability measures via a comprehensive statistical approach is required. Furthermore, the effect of trial size on a player's ranking also warrants further exploration because biomechanical screening is typically performed to identify neuromuscular and biomechanical deficits to distinguish potentially "at-risk" athletes.

2.4.4 Data analysis methods

Studies typically adopt a multi-trial procedure whereby discrete parameters (i.e., maximum or minimum values during WA) are derived by averaging data across trials (385, 484, 485). However, the averaging method (i.e., average of individual trial peaks or peak from average profile) can produce different outcome values, especially when there are temporal misalignments in the variable's waveform (116). Dames et al. (116) demonstrated angular velocity and GRF peak values from the average profile were significantly smaller ($p \leq 0.002$, ES = 0.08-0.16) compared to averaging individual trial peaks during walking gait. Unfortunately, it remains unclear in the COD biomechanical literature how researchers derive discrete data as several studies state trials were averaged (5, 39, 108, 127, 227-229, 504, 514, 516), but do not delineate whether a peak of an average profile or average of individual trial peaks method was used. Additionally, some studies fail to state whether average data was used for statistical analysis (66, 114, 115, 126, 174, 260, 301, 359, 497, 505, 574). Thus, further research into the effect of averaging methods on COD outcomes is required.

4.4.5 Velocity of centre of mass

The assessment of COM velocity during COD is of great interest to practitioners and researchers, as approach velocity is a critical factor augmenting knee joint loading (114, 290, 301, 574), but also a key determinant of faster COD performance (213, 275, 411). Single marker representation (e.g., PSIS marker, greater trochanter etc.) (114, 115, 126, 127, 140, 272-274, 358) have been previously used to estimate COM velocity during dynamic tasks; however, Vanrenterghem et al. (573) and Havens et al. (230) have expressed a note of caution when using single marker representation (i.e., PSIS or T8) or reduced marker sets (between 5th lumbar and 1st sacral vertebrae and pelvis average model) to assess COM velocity, with lower bias and greater accuracies observed with a combined lower-limb and trunk model (which accounts for 82% of BM) for estimating COM velocity.

2.4.6 Normalisation of joint moments

Joint moment and GRF data are typically normalised to reduce subject variation and permit fairer statistical comparisons between groups (369, 416). Variables are typically normalised to BM or BW, or a product of BM and height (369, 416), while allometrically scaled to BM (384), and normalising joint moment data to leg length (384, 586) are also alternative normalisation methods. A diverse range of normalisation methods have been used across the COD literature (Table 2.12); however, while the choice of normalisation method will not increase the statistical power of the data, the chosen method may influence the subsequent outcomes of the results (384). The results from previous research have shown that the normalisation method can influence the magnitude of differences between groups and influence the interpretation of an athlete's biomechanical profile (369, 416). Additionally, the seminal work of Hewett et al. (242) has been heavily scrutinised for failing to normalise joint moment and GRF data when comparing data between ACL injured and uninjured athletes (415), whereby differences of 3.6 cm and 2.4 kg in height and BM, respectively, were present. Therefore, it is integral to acknowledge and understand the implications regarding the effect normalisation has on biomechanical data outcomes when comparing data between groups, longitudinally monitoring changes in movement mechanics, and interpreting and comparing biomechanical research between studies.

Table 2.12. Summary of normalisation methods adopted within COD literature research

Normalisation methods	Studies
Body weight	(5, 114, 272-274)
Body mass	(5, 79, 98, 106, 174, 228, 229, 261, 272, 273, 275, 289, 290, 302, 346, 358, 505, 519, 574)
Body mass × Height	(105, 126, 127, 140, 189, 207, 515, 518)
Body weight × Height	(84, 93, 114, 115, 266, 267, 318, 556)
Non-normalised	(301, 370, 497)

2.4.7 Other methodological and technological considerations

Marker placement is also a critical factor for attaining accurate and reliable kinetic and kinematic data. Imprecise marker placement opens up the prospect of errors and inaccuracies in joint centre or body segment identification, thus affecting anatomical coordinate systems which may result in erroneous estimations of joint kinetic and kinematic data (40, 188, 251, 428, 523, 533, 550), as well as affecting repeatability between sessions (70, 78, 124, 201, 282, 523). Moreover, reference frames (303, 325, 501-503), direct and indirect kinematic approaches (modelling) (488, 497), and methods to estimate body segment inertial parameters (70) can also influence outcome measures, and should therefore be acknowledged when conducting 3D motion analysis and comparing research findings between studies.

2.5 Literature review summary

In summary, Chapter 2.1 highlighted that CODs are key actions associated with non-contact ACL injury, which are a serious and potentially debilitating injury. The mechanism of non-contact ACL injury has been described as multiplanar, with "high-risk" trunk and lower-limb biomechanical and neuromuscular control deficits contributing to potentially hazardous multiplanar knee joint loads and potential ACL strain. Chapter 2.2 highlighted the difficulties in screening and predicting injury risk, but underlined the importance of screening moving quality to identify "high-risk" deficits displayed by athletes to help inform future training programmes. A screening tool for evaluating cutting movement quality has been preliminary developed, but must be further developed and validated against 3D motion analysis to assist practitioners working the athletes in cutting dominant sports. Based on the review sections in Chapter 2.3, there appears to be a "performance-injury conflict" during COD; however only one study has directly assessed the determinants of performance and KAMs during COD. Consequently, further research is needed that considers both aspects to help confirm whether techniques and mechanics required for faster COD performance concurrently elevate knee joint loads. Finally, the scoping review (Chapter 2.3.5) has highlighted that COD technique modification is effective in reducing knee joint loads, but no study to date has comprehensively and additionally considered the implications for performance while containing a CG. Further research is necessary which considers both performance and injury risk aspects of COD technique modification, to help assist in the development of more effective ACL injury mitigation and COD speed programmes.

CHAPTER 3: RESEARCH METHODOLOGY

3.1 Participants

Male athletes from multidirectional sports (professional and semi-professional football clubs, cricket academies, and university sports teams) including soccer, rugby, cricket, and field-hockey were recruited. Female athletes from netball and soccer were recruited for Chapter 6 (Study 3) only. Athletes from these aforementioned sports were chosen because of the importance of COD ability (51, 156, 171, 184, 549, 601, 632) and frequency of non-contact ACL injuries in these sports (204, 248, 376, 541, 583). Inclusion criteria for each study consisted of the following: 1) minimum of five years playing the respective sport; 2) regularly performing one game and two structured skill-based sessions per week; and 3) free from lower-limb injury six months prior to testing and no history of a traumatic knee injury such as an ACL injury. The investigations were approved by the University of Salford Institutional Ethics Review Board (Appendix 1.1), and all participants were informed of the benefits and risks of the investigation, prior to signing an institutionally approved consent document to participate in the study. All participants were asked which leg they would prefer to kick a ball with to indicate limb dominance. Prior to any testing, all participants had their BM (Seca Digital Scales, Model 707, Birmingham, United Kingdom) and standing height (Stadiometer; Seca, Birmingham, United Kingdom) measured.

3.2 Change of direction speed tasks

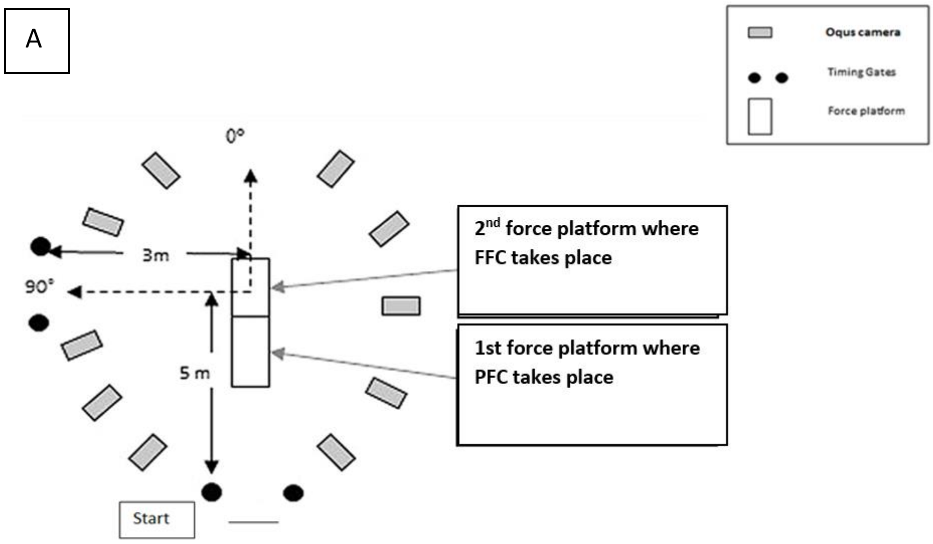
Prior to maximal COD speed tasks, all participants performed a 5-minute warm-up consisting of jogging, self-selected dynamic stretching, and familiarisation trials of the cutting task (4 per task performed submaximally at 75% of perceived maximum effort), similar to the warm-up procedures utilised in previous studies (115, 574).

For this thesis, a 70-90° side-step cutting task was investigated from both short (5-m) and long (15-m) approaches to provide an indication of low- and high-entry velocity COD ability and this was the primary angle investigated in this thesis. Schematic representations of the COD tasks are presented in Figure 3.1. Athletes perform directional changes from various approach distances, and limited research has examined the effect of approach distance/velocity on COD biomechanics; therefore, rationalising the inclusion of shorter and longer approach distances. Study 5 (Chapter 8) also included a 45˚ side-step cutting task.

All participants performed the cutting tasks in the Human Performance Laboratory on an indoor track (Mondo, SportsFlex, 10 mm; Mondo America Inc., Mondo, Summit, NJ, USA) (excluding Chapter 7; Study 4). For all tasks, participants adopted a two-point staggered stance, 0.5-m behind the start line (330), to prevent early triggering of the timing gates, and sprinted “as fast as possible” in a straight line to the cutting point before changing direction and exiting towards the finish line (Figure

3.1). Each trial was interspersed with two minutes' of rest. Completion time was measured using sets of single beam Brower timing lights (Draper, UT, USA) that were set at approximate hip height for all participants to ensure that only one body part (such as the lower torso) breaks the beam (623). Completion times for each trial were recorded to the nearest 0.001 second.

For each study (excluding Chapter 7; study 4), participants performed a minimum of six "good and acceptable" trials, while for the between-session reliability: impact of trial size study (Chapter 4), participants performed a minimum of 10 "good and acceptable trials". A "good and acceptable trial" was considered to be as follows: 1) foot contact with the 1st force platform (PFC) and second force platform (FFC) with no slide or double foot contact; 2) contact with the central portion of the last platform during FFC to ensure a homogeneous distance of travel between trials; and 3) trial performed without actively targeting the force plates (i.e., no stuttering) or premature turning. Each trial was observed by a researcher and immediately inspected using Qualisys Track Manager (QTM) software to confirm whether the trial satisfied the acceptable criteria. Trials which failed to satisfy the criteria were rejected, and another trial was performed after two minutes' of rest. All trials were performed on the right limb only (due to laboratory configuration) and were pre-planned.



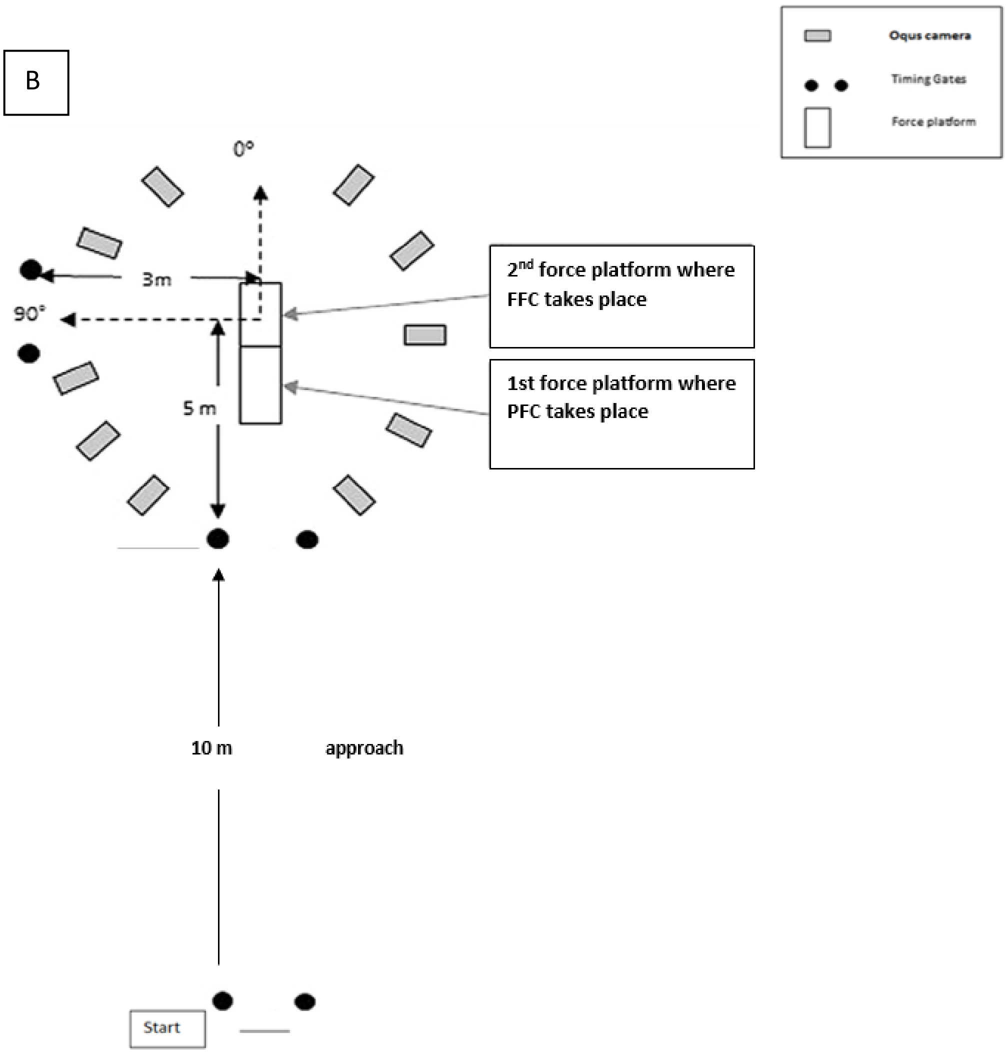


Figure 3.1. Schematic representations of the short (A) and long (B) cutting tasks

3.3 Three-dimensional motion and ground reaction force procedures and analysis

(applicable to all studies excluding chapter 7; study 4)

3.3.1 Marker placement

Data collection procedures were similar to previously published work (272, 273, 275) performed in the same laboratory (Human Performance Laboratory, G47, Mary Seacole building). Prior to the COD tasks, reflective markers (14 mm spheres) were placed bilaterally on the following body landmarks: iliac crest, anterior superior iliac spine, posterior superior iliac spine, greater trochanter, medial epicondyle, lateral epicondyle, lateral malleoli, medial malleoli, heel, fifth, second, and first metatarsal heads, acromion process, and a single marker for C7 and mid clavicle using double-sided adhesive tape (Figure 3.2). Participants also wore a 4-marker ''cluster set'' (4 retroreflective markers attached to a lightweight rigid plastic shell) on the right and left thigh and shin, which approximated the motion of these segments during the dynamic trials (Figure 3.2). This technique is suggested to reduce the influence of soft tissue artefacts and be more accurate and practical for tracking motion than individual skin markers (40, 71), with 4 markers suggested as optimal (339). The thigh and shin cluster sets were attached using Velcro-elasticated wraps. The thigh cluster sets were placed on the distal third of the thigh to reduce movement of the cluster set caused by muscle bulk and swinging of the hands. Female participants also wore an additional trunk 4-marker "cluster set" which was placed in between the scapulae (Chapter 6 only).3.3.

Imprecise marker placement opens up the prospect of errors and inaccuracies in joint centre or body segment identification, thus affecting anatomical coordinate systems and can therefore result in erroneous estimations of joint kinetic and kinematic data (40, 188, 251, 428, 523, 533, 550). Furthermore, inconsistent marker placement between testers, between sessions, and between laboratories is also problematic for repeatability (70, 78, 124, 201, 282, 523). As such, the lead researcher, who was experienced in anatomical identification and palpation, placed the markers for all participants and for each study. All participants wore Lycra shorts and standardised footwear (Balance W490, New Balance, Boston, MA, USA) to control for shoe-surface interface (152).

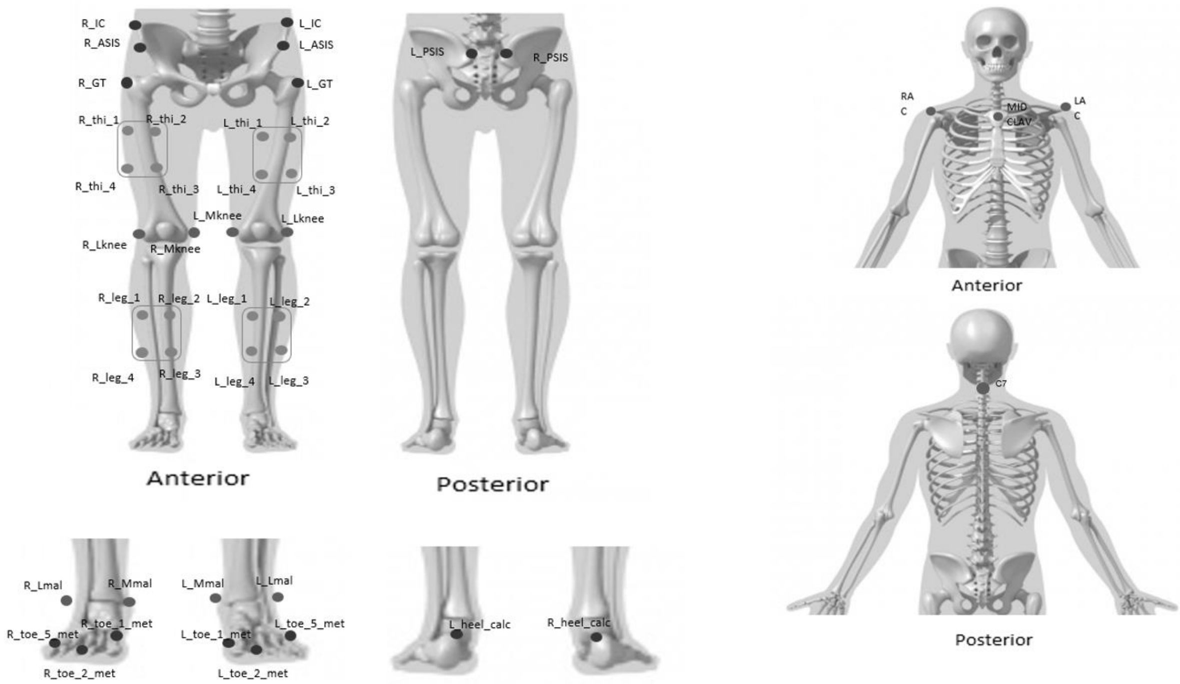


Figure 3.2. Anterior and posterior views of marker placement

3.3.2 3D motion and GRF data analysis

A capture volume for the COD tasks was created to enable data collection of the approach prior to the PFC and exit following the FFC. The capture volume was dynamically calibrated by moving a 601.7 mm wand throughout the capture area (lower-floor level to head height) (Figure 3.3) for a duration of 70 seconds with residual errors < 0.5 mm. A rigid 566 mm × 365 mm L-frame with 4 markers was also positioned on the corner of the FFC force platform for static calibration to define the global coordinate system in three planes (x, y, and z) (Figure 3.3). 3D motions of these markers were collected during the COD trials using 10 Qualisys Oqus 7 (Gothenburg, Sweden) infrared cameras (240 Hz) mounted to the ceiling and wall operating through QTM software (Qualisys, version 2.16 (Build 3520), Gothenburg, Sweden). GRFs were collected from two 600 mm × 900 mm AMTI (Advanced Mechanical Technology, Inc, Watertown, MA, USA) force platforms (Model number: 600900) embedded into the running track sampling at 1200 Hz.

From a standing trial, a six degrees of freedom kinematic model of the lower-extremity and trunk was created for each participant (Figure 3.4), including pelvis, thigh, shank, and foot using Visual3D software (C-motion, version 6.01.12, Germantown, USA) which was scaled to height and BM. This kinematic model was used to quantify the motion at the hip, knee, and ankle joints using a Cardan angle sequence x-y-z (546). The local coordinate system was defined at the proximal joint centre for each segment. The static trial position was designated as the subject's neutral (anatomical zero) alignment, and subsequent kinematic and kinetic measures were related back to this position.

Segmental inertial characteristics were estimated for each participant (128). This model utilised a CODA pelvis orientation (30) to define the location of the hip joint centre. The knee and ankle joint centres were defined as the mid-point of the line between lateral and medial markers (114, 115).

Figure 3.3. Static L-frame and wand used for calibration

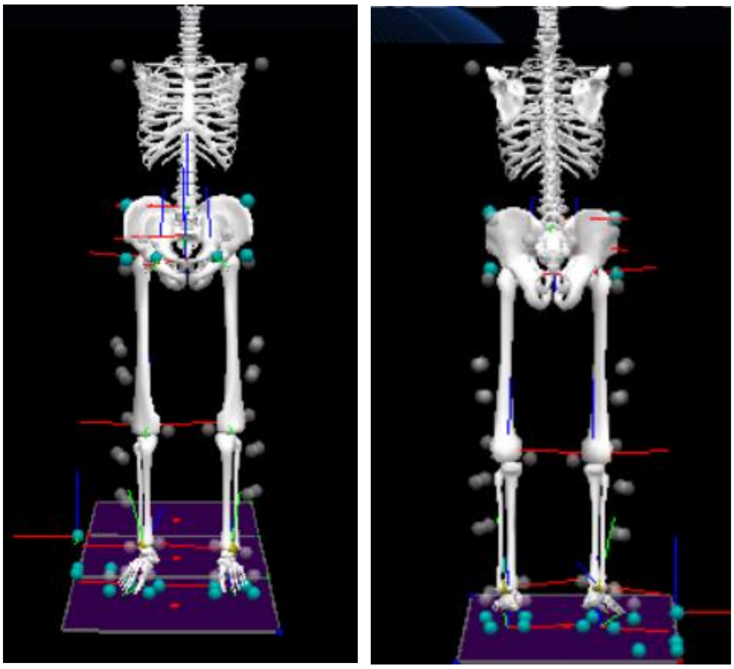


Figure 3.4. Six degrees of freedom kinematic model of the lower-extremity and trunk

Marker trajectories for all trials were labelled using QTM software before being exported as C3D files and analysed using Visual3D software. Using the pipeline function in Visual3D, joint coordinate (marker) and force data were smoothed with a fourth order Butterworth low-pass digital filter with COFs of 15 Hz and 25 Hz, based on *a priori* residual analysis (613), visual inspection of motion data, recommendations by Roewer et al. (493), and results of a pilot study presented in Appendix 2.1. Lower-limb joint moments were calculated using an inverse dynamics approach (612) through Visual3D software and were defined as external moments (i.e., external KAM will abduct the knee – distal end of tibia away from midline of body), which were normalised to BM because ACL ligament tensile strength also scales to mass (80, 415). Joint kinematics and GRFs were also calculated using Visual3D with Table 3.1 providing definitions and calculations.

The cutting trials were time normalised for each participant to 101 data points with each point representing 1% of the WA phase (0 to 100% of WA) or push-off phase (0 to 100% of push-off). Initial contact was defined as the instant of ground contact that the VGRF was higher than 20 N, and end of contact (toe-off) was defined as the point where the VGRF subsided past 20 N (272-274, 301, 304). The WA (braking) phase was defined as the instant of IC to the point of maximum knee flexion (228, 272-274), and the push-off phase (propulsive) was defined as point of maximum knee flexion to toe-off.

3.4 Kinetic and kinematic variables

A full description of variables along with definitions, abbreviations, and calculations are provided in Table 3.1. Lower-limb joint and trunk angles were calculated, while GRF braking and propulsive characteristics were normalised relative to BW, with vertical, anterior-posterior, and ML corresponding to Fz, Fx, and Fy, respectively. Similar to the procedures of Jones et al. (275), horizontal COM velocity profiles during the COD were calculated in accordance with the recommendations of Vanrenterghem et al. (573) and Havens et al. (230). Horizontal velocity of COM was calculated using a combined lower-limb and trunk model because this method is suggested to be more accurate than single marker representations (i.e., posterior superior iliac spine, greater trochanter). As such, horizontal velocity in the direction of motion of the model COM (lower-limb and trunk model) was calculated at PFC IC to determine approach velocity, FFC touch-down, and at toe-off of the FFC to determine exit velocity. Furthermore, to provide an indication of the true COD angle, an executed cutting angle was calculated in accordance with Besier et al. (39) (Table 3.1). Joint angle data were assessed at IC, peak, and ROM. Discrete point analysis was performed throughout the thesis and thus, the average of individual trial peaks for each variable were calculated as recommended by Dames et al. (116), and shown in a methodological study outlined in appendix 2.2 which has been published (142). Table 3.2 also indicates which variables were investigated for each study in the thesis.

Table 3.1. Definitions and calculations for kinetic and kinematic variables examined during cutting across studies

	Variable	Foot contact	Abbreviation	Definition or calculation
Sagittal plane joint moments	Peak ankle dorsi-flexion, knee flexion, and hip flexion moment	PFC and FFC	ADFM, KFM, HFM	Peak external joint moments during weight acceptance using inverse dynamics
Sagittal plane joint angles	Ankle dorsi-flexion angle; knee flexion angle; hip flexion angle	PFC and FFC	ADFA, KFA, HFA	Derived from the following order of rotations: flexion (+)/extension (-). Angle between two segments
Frontal plane injury risk parameters	Peak knee abduction moments	FFC	KAM	Peak external knee abduction moment (+ abduction/- adduction) during weight acceptance of FFC using inverse dynamics. Synonymous with knee valgus moment.
	Knee abduction angle	FFC	KAA	Knee abduction angle (-) during weight acceptance /adduction (+)
Transverse plane injury risk parameters	Peak knee rotation moment	FFC	KRM	Peak external knee rotation moment (+ external/- internal) during weight acceptance using inverse dynamics
	Knee rotation angle	FFC	KRA	Knee rotation angle (- internal/ + external) during weight acceptance
GRF	Peak vertical braking force (Fz)	PFC and FFC	VBF	Peak normalised VGRF (Fz) value during weight acceptance
	Mean vertical braking force (Fz)	PFC and FFC	Mean VBF	Average normalised VGRF (Fz) during weight acceptance
	Peak horizontal braking force (Fx)	PFC and FFC	HBF	Peak normalised HGRF (Fx) value during weight acceptance
	Mean horizontal braking force (Fx)	PFC and FFC	Mean HBF	Average normalised HGRF (Fx) during weight acceptance
	Peak vertical propulsive force (Fz)	FFC	VPF	Peak normalised VGRF (Fz) value during push-off
	Mean vertical propulsive force (Fz)	FFC	Mean VPF	Average normalised VGRF (Fz) during push-off
	Peak horizontal propulsive force (Fx)	FFC	HPF	Peak normalised HGRF (Fx) value during push-off
	Mean horizontal propulsive force (Fx)	FFC	Mean HPF	Average normalised HGRF (Fx) during push-off
	Peak medio-lateral braking force (Fy)	PFC and FFC	MLBF	Peak normalised MLGRF (Fy) value during weight acceptance
	Mean medio-lateral braking force (Fy)	PFC and FFC	MLBF	Average normalised MLGRF (Fy) during weight acceptance
	Peak medio-lateral propulsive force (Fy)	PFC and FFC	MLPF	Peak normalised MLGRF (Fy) value during push-off
	Mean medio-lateral propulsive force (Fy)	PFC and FFC	MLPF	Average normalised MLGRF (Fy) during push-off
	Resultant braking force	PFC and FFC	RBF	Calculated using Pythagoras theorem (resultant force = √ ((Vertical force^2) + Horizontal force^2)) over weight acceptance
	Resultant propulsive force	FFC	RPF	Calculated using Pythagoras theorem (resultant force = √ ((Vertical force^2) + (Horizontal force^2)) over push-off
	Braking force ratio	Between the two contacts	-	FFC braking force / PFC braking force
	Ground contact time	PFC and FFC	GCT	Duration from IC to toe-off
Trunk variables	Lateral trunk flexion	FFC	-	Angle of trunk relative to vertical line perpendicular to the pelvis in frontal plane: (0˚) upright / (+) medial trunk flexion away from plant foot/ (-) lateral trunk flexion towards plant foot
	Forward trunk inclination angle	PFC and FFC	-	Angle of trunk relative to a vertical line, (+) forward trunk lean/ (-) backward trunk lean
Hip, pelvis, and foot	Hip rotation angle	FFC	HRA	Femur internally rotated (-)/ external rotation (+)
	Hip abduction angle	FFC	-	Hip abduction angle (-) during weight acceptance phase of FFC/ Adduction (+)
	Pelvic rotation	FFC	-	Angle of pelvis in transverse plane relative to global coordinate system. (0˚) straight and perpendicular, (+) rotation towards intended direction of travel (-) rotation away from intended direction of travel

	Lateral foot plant distance	FFC	-	Lateral distance from initial foot contact of foot COM to proximal end of pelvis
	Initial foot progression angle	FFC	IFPA	Angle of foot progression relative to global coordinate system: straight (0°)/inward rotation (+)/outward rotation (-) angle (°)
Velocity/COM	Horizontal velocity of COM	PFC and FFC	-	The first derivative of the model COM (combined lower-limb and trunk model) position was computed to derive anterior-posterior (x), vertical (z), and ML (y) over the PFC and FFC. Resultant horizontal plane velocity was calculated using the following formula: $\sqrt{((\text{COM vel } (x)^2) + (\text{COM vel } (y)^2))}$ to provide a "velocity profile" along the path of the participants COM during the cut. Model COM velocity at PFC touch-down (approach), FFC touch-down, and toe-off in FFC (exit).
	Angle of COD (based on COM)	FFC	-	Calculated using trigonometry based on y and x displacements of model COM at toe-off ((COD angle = $\tan^{-1}(y/x)$)

Key: PFC: Penultimate foot contact; FFC: Final foot contact; COM: Centre of mass; COD: Change of direction; IC: Initial contact; GRF: Ground reaction force; VGRF: Vertical GRF; HGRF: Horizontal GRF; MLGRF: Medio-lateral GRF; vel: velocity

Table 3.2. Variables investigated for each study

	Variable	Study 1	Study 2	Study 3	Study 4	Study 5
Performance	Completion time (s)	✔	✔	✔	✔	✔
	Approach time (s)	✔	✔			
	COD deficit (s)				✔	
Qualitative screening	CMAS			✔	✔	
Sagittal plane joint moments	pk ADFM, KFM, HFM (Nm/kg)	✔ (PFC & FFC)	✔(FFC)	✔ (FFC)		✔(FFC)
Sagittal plane joint angles	ADFA (°) KFA (°) HFA (°)	✔ (pk) ✔ (pk) ✔ (pk) (PFC & FFC)	✔ (pk, IC, ROM) ✔ (pk, IC, ROM) ✔ (pk, IC, ROM) (FFC only)	✔ (pk, IC, ROM) (FFC only)		✔ (pk, IC, ROM) ✔ (pk, IC, ROM) ✔ (pk, IC, ROM) (FFC only)
Frontal plane injury risk parameters	pk KAM (Nm/kg)	✔	✔	✔		✔
	KAA (°)	✔ (pk)	✔ (pk, IC)	✔ (pk, IC, ROM)		✔ (pk, IC)
Transverse plane injury risk parameters	pk KIRM (Nm/kg)	✔	✔	✔		✔
	KRA (°)		✔ (pk, IC)	✔ (pk, IC)		✔ (pk, IC)
Braking GRF	pk VBF (BW)	✔ (FFC & PFC)	✔ (FFC & PFC)	✔ (FFC & PFC)		✔ (FFC)
	Mean VBF (BW)	✔ (FFC & PFC)	✔ (FFC & PFC)	✔ (FFC & PFC)		✔ (FFC)
	pk HBF (BW)	✔ (FFC & PFC)	✔ (FFC & PFC)	✔ (FFC & PFC)		
	Mean HBF (BW)	✔ (FFC & PFC)	✔ (FFC & PFC)	✔ (FFC & PFC)		✔ (FFC & PFC)
	RBF (BW)		✔ (FFC & PFC)			
	HBF ratio	✔	✔	✔		
Propulsive GRF	pk VPF (BW)		✔			
	Mean VPF (BW)		✔			
	pk HPF (BW)		✔			✔
	Mean HPF (BW)		✔			✔
	pk MLPF (BW)		✔			✔
	Mean MLPF (BW)		✔			✔
	pk RPF (BW)		✔			
	Mean RPF (BW)		✔			
	FFC GCT (s) PFC GCT (s)	✔ ✔	✔ ✔			✔ ✔
Trunk variables	Lateral trunk flexion angle (°)		✔ (pk, IC, ROM)	✔ (IC)		✔ (IC)
	Forward trunk inclination angle (°)		✔ (FFC & PFC) (IC)	✔ (FFC & PFC) (IC)		✔ (FFC)
Hip, pelvis, and foot	HRA (°)		✔ (IC, pk)	✔ (IC)		✔ (IC, pk)
	Hip abduction angle (°)		✔ (IC)			✔ (IC)
	Pelvic rotation (°)		✔ (IC)			✔ (IC)
	Lateral foot plant distance (m)		✔ (IC)	✔ (IC)		✔ (IC)
	Initial foot progression angle (°)		✔ (IC)	✔ (IC)		✔ (IC)
Other	Approach velocity (m/s)		✔	✔		✔
	FFC touch-down velocity (m/s)		✔			✔
	Exit velocity (m/s)		✔			✔
	Δ in PFC velocity (m/s)		✔			
	Δ in FFC velocity (m/s)		✔			
	COD angle (°)		✔			✔

Key: GCT: Ground contact time; PFC: Penultimate foot contact; FFC: Final foot contact; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFM: Ankle dorsi-flexion moment; HFA: Hip flexion angle; KFA: Knee flexion angle; ADFA: Ankle dorsi-flexion angle; RPF: Resultant propulsive force; RBF: Resultant braking force; VPF: Vertical propulsive force: VBF: Vertical braking force; HPF: Horizontal propulsive force; HBF: Horizontal braking force; MLPF: Medio-lateral propulsive force; HRA: Hip rotation angle; ROM: Range of motion; IC: Initial contact; pk: peak; COD: Change of direction; KRA: Knee rotation angle; BW: Body weight

CHAPTER 4: Between-Session Reliability of Lower-Limb Biomechanics During Cutting: Effect of Trial Size

4.1 INTRODUCTION

Changing direction is a key manoeuvre associated with non-contact ACL injuries (298, 426, 583) and also an action connected with decisive moments in sport (171, 510, 601). As such, understanding and identifying the biomechanical parameters associated with injury risk (e.g., KIRM, KAM, KAA, VGRF, and KFA) (183, 242, 298) and performance determinants (e.g., braking force, GCT, joint moments) (143, 228, 346) are of interest to practitioners working with multidirectional athletes. 3D motion analysis is commonly used to evaluate COD biomechanics; however, in order for practitioners to be confident in the collection and subsequent interpretation of data, it is central to understand the reliability and variability of the assessment (16, 27, 254). Understanding the variability and measurement error of COD biomechanical variables is fundamental for establishing "real" changes in performance (16, 254, 474). Typical sources of error during 3D motion analysis include marker placement (70, 124, 282), movement variability (task execution) (21, 22, 466, 570), and data collection, calibration, processing, and equipment errors (27, 355). Unfortunately, reliability measures are rarely reported for COD biomechanical variables (Table 4.2). Furthermore, training interventions have shown reductions in knee joint loads during COD (93, 126, 140, 292, 555, 556), but have not acknowledged measurement errors when monitoring changes (Tables 2.5-2.8), thus reducing the certainty that such changes can be interpreted as "real".

Several studies have examined the within- (5, 101, 142, 143, 205, 370) and between-session (5, 38, 346, 370, 497, 514, 515, 639) reliability of COD biomechanics, and indicate most parameters are reliable (Table 4.1). However, it is worth noting that the majority of the COD reliability studies have only used single statistical designs to assess reliability, such as an intraclass correlation coefficient (ICC) (346) or correlation of multiple coefficients (CMC) (514, 515) which are limited, because they provide no indication of the level of agreement, variability, or magnitude of differences between trials or sessions (335, 474). For a more holistic overview of reliability, it has been recommended that a combination of statistical tests (e.g., statistical tests for systematic bias, within-subject variation, and retest correlation) are used (16, 257, 335).

Only a limited number of studies have comprehensively examined the between-session reliability of COD biomechanics using several statistical tests (5, 370, 639), and, in general, report most variables such as lower-limb sagittal plane kinetics and kinematics and GRF variables as reliable (Table 4.1). However, lower-limb transverse and frontal plane kinetics and kinematics have demonstrated poorer reliability measures (5, 370, 497). Alenezi et al. (5) found transverse kinematics to display low between-session ICCs (0.42-0.51), while Mok et al. (370) expressed caution regarding peak KAM between-session ranking (ρ = 0.59), because only six of 12 athletes classified as "high-risk" (top 30% from session 1) were also classified as "high-risk" in session two. Similarly, Sankey et al. (497) reported high variability for knee joint moments between sessions during cutting (31.8, 24.1, and 16.9 Nm for sagittal, frontal, and transverse planes, respectively). Taken together, these findings are concerning because KAMs prospectively have been shown to predict injury (242), and are typically used for screening and evaluative purposes to subsequently inform future training (241, 372). The variability for KAMs suggest that different evaluations regarding an athlete's "injury risk profile" could be made. Consequently, further research quantifying the reliability of "high-risk" parameters during COD is warranted.

Table 4.1. Summary of research that has quantified between-session reliability for COD biomechanical variables

Study	Sample size	COD task (angle)	Statistical approach	Variables	Results
Sigward & Powers (515)	n = 5* soccer players	45° cut*	CMC	Sagittal, frontal, and transverse plane knee kinematics and kinetics	• Excellent to fair CMC values for sagittal, frontal, and transverse plane knee kinematics (0.98, 0.63, and 0.61). • Excellent repeatability for sagittal, frontal and transverse knee joint moments (0.93, 0.90, and 0.93).
Besier et al. (39)	11 healthy males Performed 3 sessions over 3 weeks	Sidestepping 30° and 60° - Crossover cut 30° 10 trials of each task Approach speed 3 m/s	ANOVA, r^2	Sagittal, frontal, and transverse plane knee kinetics. VGRF, MLGRF, APGRF. Knee flexion angle	• Week 1 had < repeatability in GRF and knee joint moments compared with the second and third testing sessions, with lower r^2 values. • $p < 0.05$ in KAM and IRM moments, and knee flexion angles between weeks 1–2 and weeks 1–3. – magnitude and direction not provided • No significant differences in any parameters between week 2 and week 3. • Weeks 2 and 3: GRF – Joint moments – very repeatable within- and between-sessions (r^2)
Zebis et al. (639)	8 female elite handball players Two-sessions 9±4 days apart	Side-step 2-m approach – 5 trials "as fast and forceful as possible"	ICC, Wilcoxon test, spearman's rho, CV%	Knee and hip angle at landing, peak Fz (no other details provided), GCT	• ICC = 0.851-0.922 and spearman's rho = 0.857-0.881 • CV% = 7.99-12.9% for all variables excluding knee angle CV% = 20.21% • Significantly shorter GCT (p = 0.036, ES = 0.50), knee flexion small difference (ES = 0.33), other variables trivial (ES ≤ 0.16).
Alenezi et al. (5)	15 recreational athletes Two-sessions 7 days apart at the same time of day	90° cut – 3 trials 3 sessions: 2 in same day – between-session 1 week later Perceived max velocity	ICC SEM SDD	Hip flexion angle and moment, hip adductor angle and moment, hip internal rotation angle, KAA and KAM, knee flexion angle and moment, ankle dorsi-flexion angle and moment, and VGRF	**Joint angles:** • Within-day ICC = 0.63–0.96 were higher than between-session ICC = 0.42–0.83. • Hip internal rotation angle (ICC = 0.51) and knee internal rotation angle (ICC = 0.42) demonstrated lower ICCs **Moments:** • Within-day ICC = 0.79– 0.94 were higher than between-session ICC = 0.83–0.92 **VGRF:** • Higher ICC = 0.95 within-day compared to between ICC = 0.88 **SEM values:** Moments: 0.13–0.56 Nm/kg, Joint angle: 1.73–5.15°; VGRF: 0.18-0.28 (BW)
Marshall et al. (346)	15 male Gaelic hurling	75° cut- 3 trials Two sessions 7 days apart	ICC	Completion time, GCT Kinematics: Torso,	**Kinematics** • 17 of 18 variables displayed excellent reliability (ICC > 0.75)

	Two-sessions 7 days apart at the same time of day	As fast as possible		pelvis, hip, knee, ankle, and foot ROM Kinetics: VGRF, MLGRF, APGRF. Hip, knee, and ankle: sagittal, frontal, and transverse moments, concentric and eccentric power	• Torso ROM (ICC = 0.60) **Kinetics** • 10 of 25 variables excellent (ICC > 0.75) • 13 of 25 variables fair to good (ICC 0.40–0.75) • 2 of 25 were poor (ICC <0.4) (hip abductor and knee varus moment)
Mok et al. (370)	19 handball and 22 elite football females Two-sessions on average two weeks apart	Sport-specific handball and soccer cuts 3, 4, and 5 trials	ICC, TE, spearman's ranking between-sessions, t-test, ES	Sagittal, transverse, and frontal plane hip, knee, and ankle angles at IC and peak over contact. Sagittal, transverse, and frontal plane hip, knee, and ankle peak moments over contact	• Trials 1–3, all variables showed good to excellent within-session reliability (average ICC: 0.91) • Fair to good between-session reliability (average ICC: 0.73) **TE:** Kinematics: 1.3-4.5°; Kinetics: 4.6-35.8 Nm, and GRF: 37.5-88.7 N • Moderate positive between-session rank correlation coefficients (Average: 0.72) • 31 of 32 variables displayed trivial differences between sessions – 1 variable small difference • 6 of 32 variables were significantly different but all trivial (peak hip abduction angle, peak ankle inversion angle, ankle plantar-flexion angle at IC, peak KAM, peak ankle internal rotation moment and posterior shear force)
Sankey et al. (497)	8 recreational soccer players (4 males 4 females) Two-sessions 2 days apart	45° cut 4-5 m/s Trial size not provided	SD SPM	Sagittal, transverse, and frontal plane knee angles and moments over WA	• Variability in knee kinematics low across waveforms <5° • High variability for Sagittal, frontal, and transverse knee kinetics 31.8, 24.1 and 16.9 Nm.

Key * = full details not provided; KAA: Knee abduction angle; KAM: Knee abduction moment; TE: Typical error, ICC: Intraclass correlation coefficient; CV: Coefficient of variation; SEM: Standard error of measurement; SDD: Smallest detectable difference; CMC: Correlation of multiple coefficients; CI: Confidence interval; IC: Initial contact; IRM: Internal rotation moment; ROM: Range of motion; GCT: Ground contact time; BW: Body weight; GRF: Ground reaction force; SPM: Statistical parameter mapping; SD: Standard deviation; WA: Weight acceptance; ES: Effect size; AP: Antero-posterior; ML: Medio-lateral; V: Vertical

A further issue in evaluating COD biomechanics is determining the optimal trial size. A diverse range of trial sizes have been used to examine COD biomechanics (Table 4.2) including: three, four, five, six, and 10 trials, with subsequent measures typically averaged; however, the justification of these trial sizes is not reported. To ensure that the averaged values are valid, stable, and provide reliable representations of the participant's typical movement, it is imperative that the participant performs the optimal (minimum) number of trials (200, 220, 264, 370, 374, 383, 553). Increasing trial sizes has been shown to increase statistical power (23) and improve reliability measures (257); thus, potentially justifying the use of higher trial sizes. However, an excessively high trial size may increase the susceptibly to learning effects, increase fatigue, and may decrease athlete motivation (257), and athletes may adopt a pacing strategy. Furthermore, the associated time (additional time for data collection) and financial implications must also be acknowledged with increased trial sizes (89, 181, 200, 257, 383). As such, trial size is an important methodological consideration in the design of human movement experiments.

Table 4.2. Trial sizes used in the COD literature

Trial size	Study
Three	Kristianslund et al. (301), Vanrenterghem et al. (574), Jamison et al. (267), Chaudhari et al. (84)
Four	Havens and Sigward, (227, 228), Dempsey et al. (126)
Five	Dempsey et al. (127), Brown et al. (66), Dai et al. (114, 115)
Six	Jones et al. (272-274)
Ten	Besier et al. (39), McLean et al. (358, 359, 362)

Note: Except for Besier et al. (39), all abovementioned investigations have failed to report reliability measures

To the best of the author's knowledge, Mok et al. (370) are the only researchers to examine the effect of trial size on the reliability of COD biomechanics, observing only minor improvements in ICC values (ICC = 0.91 to 0.95 within-session and 0.73 to 0.77 between-session) when increasing trial size from three to five. However, the authors only compared ICCs between trial sizes, and failed to consider its effect on measurement error. Additionally, Mok et al. (370) did not compare outcome values between trial sizes, and failed to examine the stability of the mean, using the SET as adopted in other trial size investigations for evaluating dynamic movement (200, 220, 264). Therefore, it is uncertain whether systematic differences (bias) between trial sizes were present (i.e., potential learning, motivation, or fatiguing effects) (16, 257, 335). Furthermore, Mok et al. (370) only considered increasing trial size from three to five, although some researchers have used 10 trials for evaluating COD biomechanics (Table 4.2); this therefore, warrants further investigation.

The aims of this study, therefore, were three-fold: 1) to examine the between-session reliability of lower-limb biomechanical variables during cutting while considering the effect of trial size (three, five, eight, and ten) on reliability measures; 2) to compare outcome values between trial sizes to explore differences in values; and 3) to establish the stability of the mean for variables using the SET technique. It was hypothesised that high and acceptable reliability would be observed for cutting lower-limb biomechanical variables between sessions across all trial sizes. Additionally, it was also hypothesised that there would be no significant or meaningful differences in outcome values and reliability measures between trial sizes.

4.2 METHODS

4.2.1 Research design

A within-subject, repeated-measures research design was adopted, whereby ten trials of a 90° cutting task was performed on two separate testing sessions, seven days apart, to examine between-session reliability. This study examined the within- and between-session reliability of cutting lower-limb biomechanical variables, while determining the effect of trial size on reliability measures and outcome values. Participants were asked to replicate their fluid and food intake 24 hours before each day of testing, and to avoid strenuous exercise for 48 hours before testing. All testing was performed at the same time of day between sessions to minimise the effect of circadian rhythm (154).

4.2.2 Participants

Ten male soccer players (five amateur and five semi-professional; age: 22.9 ± 3.5 years, mass: 75.5 ± 9.6 kg, height: 1.78 ± 0.03 m, soccer training history: 13 ± 3 years) participated in this study, whereby lower-limb kinetic and kinematic data were collected over the PFC and FFC during a 90° cut (Figure 3.1b). This sample size was similar to previous COD biomechanics between-session reliability research (38, 497). For inclusion in the study, all athletes had played their respective sport for a minimum of five years and regularly performed one game and two structured skill-based sessions per week. All athletes were free from injury and none of the athletes had suffered prior traumatic knee injury such as an ACL injury. At the time of testing, players were in-season (competition phase) and stated right limb dominance (preferred kicking leg). The investigation was approved by the Institutional Ethics Review Board (HSR1617-02), and all participants were informed of the benefits and risks of the investigation, prior to signing an institutionally approved consent document to participate in the study.

4.2.3 Procedures

For a detailed description of the warm-up, long 90° cut, marker placement, and 3D motion analysis procedures, please refer to Chapter 3.2 and 3.3 in the research methodology. In brief, each participant performed ten acceptable trials of the long 90° cut each session as "fast as possible". Marker and force data were collected over the PFC and FFC, and completion time was also measured for each trial. 10 trials were used in the analysis of each participant based on visual inspection of the motion files (370).

4.2.4 Lower-limb kinetic and kinematic variables

Table 3.2 presents the dependent variables. Peak lower-limb sagittal plane joint angles and joint moments were evaluated over WA during the PFC and FFC. KAA, KAMs, and KIRMs were calculated over FFC WA. Braking force-time characteristics included peak and average vertical (Fz) and horizontal

(Fx) GRFs for the PFC and FFC over WA, and GCT was also calculated. This study quantified lower-limb biomechanical parameters that are used commonly for injury risk screening of athletes, such as peak KAMs, KAAs, KIRMs, KFAs, and VGRFs (183, 241, 372), because these discrete parameters are prospectively associated with non-contact ACL injury (242, 321) and commonly observed characteristics of injury during COD (241, 298, 426). Additionally, the reliability of sagittal plane joint angles and moments, and GRF variables were also quantified because these parameters are commonly used to evaluate deceleration mechanics over WA (5, 227, 272-274, 301, 370), and, as such, are of particular interest to biomechanists.

4.3 STATISTICAL ANALYSES

Four trial size conditions (independent variable) were explored in this study: an average of three (1-3), five (1-5), eight (1-8), and ten (1-10) trials. This was examined for each dependent variable to explore within- and between-session reliability and to explore differences in outcome values between trial sizes. The average of individual trial peaks for each variable were calculated (116, 142). All statistical analyses were performed in SPSS v 24 (SPSS Inc., Chicago, IL, USA) and Microsoft Excel (version 2016, Microsoft Corp., Redmond, WA, USA). Normality was inspected for all variables using a Shapiro-Wilks test.

4.3.1 Within- and between-session reliability

Within-session and between-session reliability were assessed using ICCs (two-way mixed effects, average measures, absolute agreement), coefficient of variation (CV%), and SEM, similar to previously used statistical procedures (370, 639). 95% confidence intervals (CI) for ICC and CV% were also calculated. The CV% was calculated as SD/mean × 100 for each participant and then averaged across all participants. The SEM was calculated using the formula: SD(pooled) * $\sqrt{(1-\mathrm{ICC})}$ (554), whereas the SDD was calculated between sessions, from the formula: (1.96 * ($\sqrt{(2)}$) * SEM (306). ICCs were interpreted based on the following scale presented by Koo and Li (299): poor (< 0.50), moderate (0.50-0.75), good (0.75-0.90), and excellent (> 0.90). Minimum acceptable reliability was determined with an ICC >0.7 and CV% <15% (28, 214). Paired sampled t-tests were used to compare lower-limb kinetics and kinematics between sessions for parametric data, while Wilcoxon-sign ranked tests were used for non-parametric data (16, 257). To explore the magnitude of differences between sessions and between trial sizes, percentage differences and effects sizes were calculated and interpreted as trivial (≤ 0.19), small (0.20 – 0.59), moderate (0.60 – 1.19), large (1.20 – 1.99), very large (2.00 – 3.99), and extremely large (≥ 4.00) (255). Cohen's d effect sizes were initially calculated and corrected using Hedges' g (231) as outlined in Figure 4.1.

$$d = \frac{M_A - M_B}{\sigma}$$

Hedges' *g* is an unbiased version of Cohen's *d*.

$$g = d \times \left(1 - \frac{3}{4(n_1+n_2)-9}\right)$$

Figure 4.1. Equations for Cohen's *d* and Hedges' *g* effect size (M: Mean, A = Group 1; B = Group 2; n = sample size; σ = pooled SD)

4.3.2 Outcome value comparisons between trial sizes

Dependent variables were compared across the four trial sizes within- and between-sessions, using a repeated measures analysis of variance (RMANOVA), with Bonferroni post-hoc pairwise comparisons in cases of significant differences for parametric variables. For non-parametric variables, a Friedman's test was used, and in cases of significant differences, individual Wilcoxon-sign ranked tests were used to explore differences. As described above, effect sizes and percentage differences were also calculated to explore the magnitude of differences. Statistical significance was defined as $p \leq 0.05$ for all tests.

4.3.3 Stability of mean (SET - minimum trial size)

The SET was employed to determine the point of mean stability (i.e., minimum trial size), as described and adopted by previous studies (200, 220, 553), and was modified in line with Gore et al. (200). This technique involved using a cumulative mean (moving point mean - adding one trial at a time) and a predefined bandwidth of the 1-10 trial SEM (i.e., 10 trial mean), to establish real differences. Stability was determined when the cumulative mean fell within the bandwidth and remained there for all subsequent trials (Figure 4.2) (200, 220, 553). The modified SET technique was used for all variables, sessions 1 and 2.

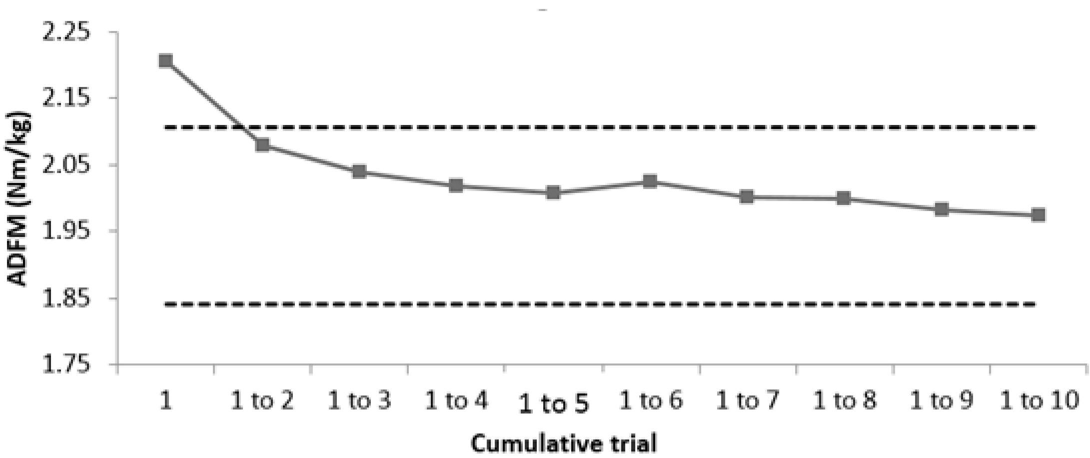


Figure 4.2. SET outlining stability in cumulative mean in reference to 10 trial SEM bandwidth

4.4 RESULTS

4.4.1 Within-session reliability

Appendix 3.1 presents the within-session reliability cutting (session 1 and 2) data including mean, SD, ICC, CV%, and SEM. Average ICCs were high for all trial sizes for sessions 1 (ICC ≥ 0.884) and 2 (ICC ≥ 0.936), with only minor improvements in ICC values as trial size increased (Appendix 3.1) and slight improvements in ICCs for session 2. Notably, all variables for all trial sizes and sessions achieved minimum acceptable reliability criteria and displayed moderate to excellent ICCs (ICC ≥ 0.731) (Appendix 3.1). Average CV% was similar between trial sizes for sessions 1 (CV% = 12.8-13.5) and 2 (CV% = 10.2-12.4), but tended to slightly increase as trial size increased, and was slightly lower for session 2 (Appendix 3.1). The CV% ranged from 4.4-49.5% for session 1, with seven out of 25 variables (FFC and PFC HFM, PFC ADFM, KAA, KAM, KIRM, and PFC peak HBF) (Appendix 3.1) demonstrating unacceptable variability for trial size three, while eight out of 25 variables (FFC and PFC HFM and ADFM, KAA, KAM, KIRM and PFC peak HBF) demonstrated unacceptable variability for all other trial sizes. For session 2, the CV% ranged from 3.0-48.6%, with five out of 25 (KAA, KAM, KIRM, PFC ADFM and HFM) demonstrating unacceptable variability for trial size 3, while seven out of 25 (KIRM, KAA, KAM, PFC and FFC ADFM and HFM) variables demonstrated unacceptable variability for all other trial sizes.

4.4.2 Between-session reliability

Tables 4.3 and 4.4 present the cutting between-session reliability data including mean, SD, ICC, CV%, SEM, SDD%, *p* values, effect sizes, and percentage differences, while 95% CIs are presented in Appendix 3.2. Between-session average ICCs were high for all trial sizes (ICC = 0.818-0.885), with minor improvements in ICC values as trial size increased (Tables 4.3 & 4.4). Notably, most variables for all trial sizes and sessions demonstrated moderate to excellent ICCs, achieving minimum acceptable reliability criteria (ICC ≥ 0.732, Table 4.3); however, FFC GCT demonstrated poor to moderate ICCs, failing to meet minimum acceptable reliability criteria for all trial sizes (ICC ≤ 0.569), while KIRM and PFC ADFM demonstrated unacceptable ICC criteria (poor and moderate) (ICC = 0.414-0.653) for trial size 3 only (Table 4.3). Average CV% (8.2-11.0%) was low but decreased as trial size increased (Table 4.3). The CV% ranged from 2.4-45.0% between sessions, with six out of 25 variables (FFC and PFC HFM, PFC ADFM, KAA, KAM, KIRM) demonstrating unacceptable variability for trial size 3, while decreasing to four out of 25 variables (FFC and PFC HFM, KIRM, and KAA) for trial size 5, three variables (FFC and PFC HFM, and KAA) for trial size 10, and four variables (FFC and PFC HFM, KAA, and KAM) demonstrating unacceptable variability for trial size 8 (Table 4.3).

Comparisons between session 1 and 2 means, revealed two of 25 variables were significantly different (FFC KFM and FFC GCT) between sessions for trial size three, and three of 25 variables were statistically significantly different between sessions for trial sizes five (FFC KFM, KAM, and FFC GCT), eight (KAM, FFC GCT and FFC mean HBF), and ten (FFC KFM, KAM, and FFC GCT) (Table 4.4). Effect sizes indicated 14 variables displayed small (or greater) differences between sessions for trial size 3, while this decreased to 11 for trial sizes 5 and 8, and 10 for trial size 10 (Table 4.4). However, average percentage differences in means between sessions were low ranging from 3.8-4.9% across trial sizes, with five variables (FFC and PFC HFM, KIRM, KAA, KAM) displaying a percentage difference ≥ 10% for trial size 3, while three variables (PFC HFM, KAA, KAM) displayed a percentage difference ≥ 10% for trial sizes 5, 8, and 10 (Table 4.4).

4.4.3 Completion and approach times

Within-session completion times and approach times achieved minimum acceptable reliability criteria across all trial sizes for cutting (ICC = 0.947-0.993, CV% = 1.7-3.0). High ICCs and low CV% were observed across all trial sizes for cutting (ICC = 0.730-0.890, CV% = 2.2-3.5) between-session completion times. In addition, although not significantly different, slightly longer completion times were observed across all trial sizes for session 1 compared to session 2 for cutting ($p \geq 0.204$, $g = 0.20$-0.37, 0.3-2.8%); however, these differences were classed as small with low percentage differences. Similarly, high ICCs and low CV% were observed across all trial sizes for cutting between-session approach times (ICC = 0.958-0.973, CV% = 1.3-1.6). Although not significantly different, slightly longer approach times were observed across all trial sizes for session 1 compared to session 2 for cutting ($p \geq 0.089$, $g = 0.13$-0.17, 0.2-1.6%); however, these differences were classed as trivial with low percentage differences.

4.4.4 Stability of mean

The number of trials required to provide a representative mean across all cutting biomechanical variables, ranged from one to six, depending on the joint, variable, and anatomical plane (Appendix 3.3). On average, SET analysis revealed for session 1, 1.6 ± 1.3 trials were needed to attain stability in the mean across all variables; however, this changed for session 2 whereby a slightly lower number of trials were required 1.3 ± 0.7. Nevertheless, these results indicate ~2 trials are needed to attain stability in the mean for cutting biomechanical variables.

Table 4.3. 90˚ cut between-session reliability measures across trial sizes

Foot contact	Variable	Trial size 3 (1-3)				Trial size 5 (1-5)				Trial size 8 (1-8)				Trial size 10 (1-10)			
		ICC	**CV%**	**SEM**	**SDD%**	**ICC**	**CV%**	**SEM**	**SDD%**	**ICC**	**CV%**	**SEM**	**SDD%**	**ICC**	**CV%**	**SEM**	**SDD%**
FFC	PK ADFA (˚)	0.897	3.9	2.7	9.8	0.958	2.4	1.7	6.2	0.959	2.6	1.7	6.2	0.967	2.4	1.5	5.6
	PK KFA (˚)	0.719	6.1	3.6	16.9	0.704	5.3	3.7	17.4	0.750	4.8	3.4	16.0	0.787	4.0	3.1	14.5
	PK HFA (˚)	0.804	11.3	4.6	28.6	0.823	10.3	4.5	27.8	0.862	9.4	4.0	25.1	0.859	9.4	4.0	25.2
	PK ADFM (Nm/kg)	**0.414**	14.2	0.27	43.3	0.707	11.9	0.19	31.2	0.842	8.6	0.14	22.3	0.846	7.4	0.14	21.5
	PK KFM (Nm/kg)	0.911	5.7	0.17	12.9	0.900	5.5	0.17	13.5	0.918	4.5	0.14	11.0	0.932	4.0	0.13	10.1
	PK HFM (Nm/kg)	0.704	**16.5**	0.38	45.5	0.727	**15.0**	0.37	44.1	0.756	**15.7**	0.36	42.3	0.760	**15.9**	0.36	41.5
	PK KAA (˚)	0.915	**45.0**	1.7	73.4	0.929	**35.6**	1.7	65.3	0.927	**36.0**	1.7	65.6	0.938	**32.1**	1.6	60.2
	PK KAM (Nm/kg)	0.880	**16.1**	0.12	35.9	0.904	14.5	0.12	32.8	0.909	**15.2**	0.12	32.5	0.920	13.1	0.11	31.2
	PK KIRM (Nm/kg)	**0.653**	**19.9**	0.15	55.6	0.778	**17.2**	0.13	45.6	0.899	13.9	0.10	34.1	0.892	13.7	0.10	34.5
	Mean HBF (BW)	0.872	7.2	0.06	20.7	0.903	5.9	0.05	16.3	0.892	5.4	0.05	15.8	0.905	5.3	0.05	14.2
	Mean VBF (BW)	0.927	5.6	0.08	14.0	0.947	5.2	0.07	11.5	0.952	4.6	0.06	10.4	0.963	3.9	0.05	9.0
	PK HBF (BW)	0.977	6.1	0.07	13.8	0.978	5.3	0.07	13.0	0.981	4.6	0.06	11.5	0.980	4.6	0.06	11.5
	PK VBF (BW)	0.916	8.3	0.20	21.8	0.928	7.2	0.18	19.4	0.943	6.4	0.15	16.6	0.938	7.1	0.16	17.3
PFC	PK ADFA (˚)	0.787	6.0	4.4	15.3	0.842	5.0	3.5	12.2	0.826	5.2	3.5	12.1	0.837	5.2	3.4	11.9
	PK KFA (˚)	0.926	3.4	2.9	7.7	0.955	2.6	2.3	6.2	0.961	2.6	2.1	5.6	0.960	2.6	2.1	5.7
	PK HFA (˚)	0.920	3.8	1.9	9.4	0.916	3.7	1.9	9.6	0.892	4.3	2.2	10.7	0.873	4.3	2.3	11.5
	PK ADFM (Nm/kg)	0.725	**20.5**	0.14	52.8	0.872	15.0	0.09	35.5	0.905	11.2	0.08	29.2	0.925	10.0	0.07	26.3
	PK KFM (Nm/kg)	0.843	8.7	0.26	20.5	0.843	8.2	0.27	21.6	0.864	7.0	0.26	20.5	0.878	6.5	0.24	19.1
	PK HFM (Nm/kg)	0.702	**24.9**	0.47	75.6	0.711	**23.1**	0.44	70.0	0.780	**20.9**	0.36	57.2	0.792	**19.9**	0.33	53.1
	Mean HBF (BW)	0.929	7.0	0.04	18.7	0.929	7.6	0.04	19.1	0.950	6.8	0.03	16.5	0.955	6.3	0.03	15.6
	Mean VBF (BW)	0.949	6.2	0.05	14.3	0.963	5.9	0.04	12.6	0.978	4.4	0.03	9.8	0.974	5.1	0.04	10.6
	PK HBF (BW)	0.877	14.6	0.17	34.1	0.907	13.5	0.15	30.0	0.934	11.9	0.13	26.4	0.946	10.5	0.12	23.8
	PK VBF (BW)	0.905	8.9	0.18	21.7	0.947	6.4	0.13	15.8	0.973	5.7	0.10	11.7	0.966	5.9	0.11	13.1
GCT	FFC GCT (s)	**0.334**	6.6	0.025	24.6	**0.314**	6.6	0.026	24.8	**0.440**	5.9	0.022	21.5	**0.569**	5.1	0.018	17.9
	PFC GCT (s)	0.788	7.9	0.013	17.6	0.791	6.8	0.013	17.2	0.737	7.5	0.014	19.1	0.774	6.8	0.013	17.3
	Average	0.818	11.0			0.850	9.5			0.872	8.8			0.885	8.2		

Key: ADFM: Ankle dorsi-flexion moment; KFM: Knee flexion moment; HFM: Hip flexion moment: ADFA: Ankle dorsi-flexion angle; KFA: Knee flexion angle; HFA: Hip flexion angle; FFC: Final foot contact; PFC: Penultimate foot contact; PK: Peak; GRF: Ground reaction force; KAA: Knee abduction angle; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; HBF: Horizontal braking force; VBF: Vertical braking force; GCT: Ground contact time; ICC: Intraclass correlation coefficient; CV%: Coefficient of variation; SEM: Standard error of measurement; SDD%: Smallest detectable difference; Bold denotes failed to achieve minimum acceptable reliability criteria

Table 4.4. 90˚ cut between-session reliability and descriptives

Foot contact	Variable	Trial size 3 (1-3)					Trial size 5 (1-5)					Trial size 8 (1-8)					Trial size 10 (1-10)				
		Mean	SD	*p*	*g*	% diff	Mean	SD	*p*	*g*	% diff	Mean	SD	*p*	*g*	% diff	Mean	SD	*p*	*g*	% diff
FFC	PK ADFA (˚)	75.8	8.0	0.222	-0.24	-2.8	75.5	8.1	0.799	-0.04	-0.4	76.0	8.2	0.253	-0.15	-1.7	76.1	8.3	0.878	0.00	0.0
	PK KFA (˚)	59.5	6.0	0.448	-0.23	-2.8	59.1	6.0	0.317	-0.29	-3.5	58.8	6.1	0.467	-0.22	-2.7	58.6	6.0	0.387	-0.23	-2.8
	PK HFA (˚)	45.0	9.6	0.278	0.28	6.6	44.7	9.8	0.587	0.12	3.1	43.9	10.0	0.457	0.19	4.5	43.6	9.8	0.597	0.12	3.0
	PK ADFM (Nm/kg)	1.72	0.28	0.514	-0.25	-5.6	1.69	0.31	0.757	-0.07	-1.6	1.73	0.32	0.830	-0.07	-1.5	1.76	0.32	0.972	-0.01	-0.2
	PK KFM (Nm/kg)	3.55	0.54	**0.031***	-0.36	-6.1	3.52	0.52	**0.008***	-0.41	-6.2	3.53	0.48	0.055	-0.35	-5.8	3.51	0.48	**0.012***	-0.36	-5.4
	PK HFM (Nm/kg)	2.35	0.63	0.157	0.43	12.8	2.35	0.63	0.372	0.25	7.8	2.34	0.65	0.346	0.27	8.4	2.37	0.65	0.508	0.25	7.7
	PK KAA (˚)	6.5	5.7	0.126	0.26	21.9	7.0	6.0	0.074	0.28	22.9	7.3	6.2	0.074	0.27	21.8	7.3	6.2	0.066	0.25	20.5
	PK KAM (Nm/kg)	0.96	0.34	0.066	0.42	15.1	0.99	0.37	**0.012***	0.48	17.8	0.99	0.38	**0.028***	0.42	15.3	0.99	0.39	**0.028***	0.40	15.4
	PK KIRM (Nm/kg)	-0.74	0.22	0.100	0.56	17.4	-0.77	0.24	0.352	0.26	8.9	-0.79	0.29	0.262	0.22	8.3	-0.80	0.29	0.292	0.22	7.9
	Mean HBF (BW)	0.86	0.17	0.053	-0.40	-9.0	0.87	0.16	0.082	-0.33	-6.3	0.88	0.15	**0.043***	-0.36	-7.4	0.89	0.14	0.058	-0.34	-6.0
	Mean VBF (BW)	1.64	0.30	0.273	-0.18	-3.6	1.64	0.29	0.257	-0.15	-2.8	1.64	0.27	0.194	-0.18	-3.5	1.65	0.27	0.238	-0.14	-2.5
	PK HBF (BW)	1.42	0.46	0.957	-0.01	-0.2	1.43	0.45	0.600	-0.05	-1.5	1.43	0.42	0.858	-0.02	-0.6	1.44	0.42	0.506	-0.06	-1.9
	PK VBF (BW)	2.59	0.68	0.399	0.15	4.2	2.57	0.65	0.443	0.11	3.0	2.56	0.62	0.490	0.12	3.1	2.58	0.63	0.340	0.15	3.8
PFC	PK ADFA (˚)	79.8	8.6	0.501	-0.18	-2.3	79.5	8.2	0.897	-0.03	-0.3	79.5	7.6	0.536	-0.15	-1.7	79.1	7.8	0.978	-0.01	-0.1
	PK KFA (˚)	104.0	10.2	0.438	0.13	1.4	104.3	10.7	0.219	0.15	1.6	104.4	10.5	0.408	0.11	1.2	104.1	10.4	0.397	0.11	1.1
	PK HFA (˚)	55.4	6.3	0.948	0.01	0.2	55.7	6.4	0.874	0.03	0.4	55.9	6.2	0.959	0.04	0.4	55.9	6.1	0.957	0.01	0.2
	PK ADFM (Nm/kg)	0.71	0.23	0.603	-0.15	-6.0	0.72	0.24	0.409	-0.15	-5.6	0.72	0.23	0.368	-0.19	-7.4	0.71	0.24	0.715	-0.06	-2.1
	PK KFM (Nm/kg)	3.54	0.62	0.106	0.37	7.0	3.52	0.64	0.367	0.20	4.1	3.51	0.66	0.139	0.22	4.4	3.51	0.66	0.399	0.18	3.6
	PK HFM (Nm/kg)	1.73	0.76	0.319	-0.30	-17.2	1.73	0.72	0.297	-0.28	-13.9	1.74	0.69	0.301	-0.31	-16.7	1.70	0.65	0.386	-0.32	-15.2
	Mean HBF (BW)	0.56	0.14	0.765	0.05	1.3	0.55	0.14	0.723	0.05	1.4	0.56	0.14	0.680	0.07	1.8	0.56	0.15	0.655	0.06	1.7
	Mean VBF (BW)	0.97	0.22	0.591	0.08	1.8	0.96	0.22	0.617	0.05	1.2	0.96	0.23	0.561	0.07	1.7	0.96	0.22	0.372	0.09	2.2
	PK HBF (BW)	1.39	0.46	0.646	0.10	3.5	1.39	0.47	0.445	0.15	5.5	1.40	0.50	0.386	0.16	5.8	1.41	0.51	0.333	0.16	6.0
	PK VBF (BW)	2.36	0.57	0.826	-0.04	-1.1	2.36	0.57	0.635	0.05	1.3	2.36	0.60	0.761	0.04	1.1	2.38	0.60	0.404	0.10	2.6
GCT	FFC GCT (s)	0.285	0.024	0.308	0.42	4.6	0.286	0.024	0.512	0.25	2.6	0.282	0.023	0.423	0.33	3.6	0.280	0.023	0.479	0.25	2.5
	PFC GCT (s)	0.203	0.027	**0.012***	0.64	8.8	0.204	0.026	**0.032***	0.61	8.2	0.202	0.025	**0.024***	0.57	7.8	0.200	0.025	**0.027***	0.59	7.8

Key: ADFM: Ankle dorsi-flexion moment; KFM: Knee flexion moment; HFM: Hip flexion moment: ADFA: Ankle dorsi-flexion angle; KFA: Knee flexion angle; HFA: Hip flexion angle; FFC: Final foot contact; PFC: Penultimate foot contact; PK: Peak; GRF: Ground reaction force; KAA: Knee abduction angle; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; HBF: Horizontal braking force; VBF: Vertical braking force; GCT: Ground contact time; ES: Effect size; ICC: Intraclass correlation coefficient; CV%: Coefficient of variation; SD: Standard deviation; *: $p \le 0.05$; **: $p \le 0.001$

Trivial ES (≤ 0.19)	Small ES (0.20-0.59)	Moderate ES (0.60-1.19)	Large ES (1.20-1.99)	Very Large ES (2.00-3.99)	Extremely Large ES (≥4.00)

4.4.5 Comparisons in outcome values between trial sizes

Outcome value comparisons between trials sizes for session 1, 2, and between-session data are presented in Appendix 3.3, which contain *p* values, ESs, and % differences. For session 1, 2, and between-sessions, RMANOVA or Friedman's test revealed trial size had a significant effect ($p \leq 0.05$) on three (FFC HFA, KAA, and FFC GCT), one (KAA), and two (FFC HFA and KAA) out of 25 variables, respectively. Pairwise comparisons are presented in Appendix 3.3 for session 1, 2 and between-session data, but in most cases, trial size had a non-significant, minimal, and trivial effect on outcome values.

4.5 DISCUSSION

This is the first study to comprehensively examine the within- and between-session reliability of 90˚ cutting biomechanics, while considering the effect of trial size (up to 10 trials), while also examining the reliability of PFC braking characteristics. This study quantified lower-limb biomechanical parameters that are used commonly for injury risk screening and risk stratifying athletes, such as peak KAMs, KAAs, KIRMs, KFA, and VGRF (183, 241, 372), because these discrete parameters are prospectively associated with non-contact ACL injury (242, 321) and commonly observed characteristics of injury during COD (241, 298, 426). Additionally, the reliability of sagittal plane joint angles and moments, and GRF variables were also quantified because these parameters are commonly used to evaluate deceleration mechanics over WA (5, 227, 272-274, 301, 370), and, as such, are of particular interest to biomechanists.

Average ICCs were high within- and between-sessions for cutting variables (Tables 4.3 & 4.4, Appendix 3.1), in line with the ICC values observed by previous researchers (Table 4.1) (5, 101, 143, 205, 346, 370, 639); however, ICC values were slightly higher for session 2 and CV% was slightly lower. No statistically significant differences in approach times and completion times were observed (≤ 3%); therefore, the improved ICCs and CV% for session 2 are mostly likely a result of increased familiarisation with the task and a potentially more consistent movement strategy. Nevertheless, most variables within- and between-sessions achieved minimum acceptable ICC reliability criteria (Tables 4.3 & 4.4, Appendix 3.1). Interestingly, no substantial differences in within- and between-session ICC values between trials sizes were observed, with only minor improvements in ICC values as trial size increased (Tables 4.3, Appendix 3.1); substantiating Mok et al's (370) findings and the study hypothesis. Additionally, the present study is one of a limited number of studies to calculate the SEM and CV% of cutting biomechanical variables (Table 4.1). Alenezi et al. (5) reported SEM values ranging from 0.13–0.56 Nm/kg and 1.73–5.15˚ for joint moment and angle data, respectively, during a 90˚ cut in recreational athletes. Mok et al. (370) reported TEs of 1.3-4.5˚, 4.6-35.8 Nm, and 37.5-88.7 N for cutting lower-limb joint kinematics, kinetics, and GRF data, while Sankey et al. (497) also documented variability of < 5˚

for knee kinematics between sessions. These aforementioned measurement error values were similar to the SEM values observed in the present study for male soccer athletes (Table 4.3, Appendix 3.1), and importantly, the SEM values did not vary substantially between trial sizes.

To the best of the author's knowledge, Zebis et al. (639) are the only other researchers to calculate the CV% during cutting between sessions, though this was limited to only four biomechanical variables (CV% = 7.99-20.21) (Table 4.1). The average CV% observed in the present study was considered acceptable for within- and between-sessions cutting data (Table 4.3, Appendix 3.1); however, similar to the ICC values, CV% generally improved for session 2 (Appendix 3.1). In contrast to ICC measures, minor improvements in CV% were observed with lower trial sizes and tended to increase as trial sizes increased for within-session reliability measures, but the CV% did not substantially differ between trial sizes (Table 4.3, Appendix 3.1). Notably, KAAs, KAMs, KIRMs, and PFC and FFC ADFM and HFMs, generally demonstrated high and unacceptable levels of variability across trial sizes (Table 4.3, Appendix 3.1). Conversely, in most cases, PFC and FFC sagittal plane joint angles, GRFs, and KFMs demonstrated acceptable and lower variability within- and between-sessions for all trial sizes (Table 4.3, Appendix 3.1).

Variability is expected for discrete biomechanical parameters due to sources of error such as marker placement (70, 124, 282), movement variability (task execution) (21, 22, 466, 570), and data collection, calibration, processing, and equipment errors (27, 355). Marshall et al. (346) observed poor reliability measures for hip abductor and knee varus moments, and partially attributed this to the support moment theory of Winter (611). Marshall et al. (346) stated: "The support theory suggests that once the algebraic sum of moments at the lower-extremity joints is sufficient for the activity in question, individual joint moments can vary largely." Furthermore, hip moments may be susceptible to increased error due to being more sensitive to segmental acceleration because the acceleration of the foot, shank, and thigh segments are taken into account when calculating hip joint moments (119). Thus, based on the high and unacceptable CV% values for KAA, KAM, KIRM, ADFM, and HFM, irrespective of trial size, cautious interpretation of these variables is recommended when screening and longitudinally monitoring changes in these variables.

Trial size is an important methodological consideration when evaluating movement due to its effects on reliability, stability, and statistical power (23, 200, 220, 257, 264, 370, 374, 383, 553). Therefore, it is integral that participants perform the optimal number of trials to represent the participant's typical movement (264). Mok et al. (370), to the best of the author's knowledge, is the only other study to consider the effect of trial size on cutting biomechanics reliability, observing only minor improvements in ICCs when increasing trial sizes from three to five. The current study expanded on the work of Mok et al. (370) by examining a greater trial size, comparing outcome values between trial sizes, and examining the stability in the mean using SET as adopted by other trial size investigations (200, 220, 264). Overall, there were no substantial differences in ICC, CV%, SEM, and SDD% values between trial sizes for cutting reliability data (Table 4.3, Appendix 3.1) and, in most cases, trial size had a non-significant, minimal, and trivial effect when comparing outcome values between trial sizes for within- and between-session data (Appendix 3.3); supporting the study hypothesis. Furthermore, SET analysis revealed ~2 trials were needed to attain stability in the mean for cutting biomechanical variables (Appendix 3.3). Consequently, based on the results of this study, three to five trials (good trials that meet acceptable criteria) are recommended for examining lower-limb cutting biomechanics, with no additive benefit or apparent requirement for greater trial sizes. Using lower trial sizes of three or five may reduce the time required for data collection and analysis for one subject, reduce financial costs due to less time for testing and analysis, and reduce the fatigue and potential injury risk for participants during maximal effort testing.

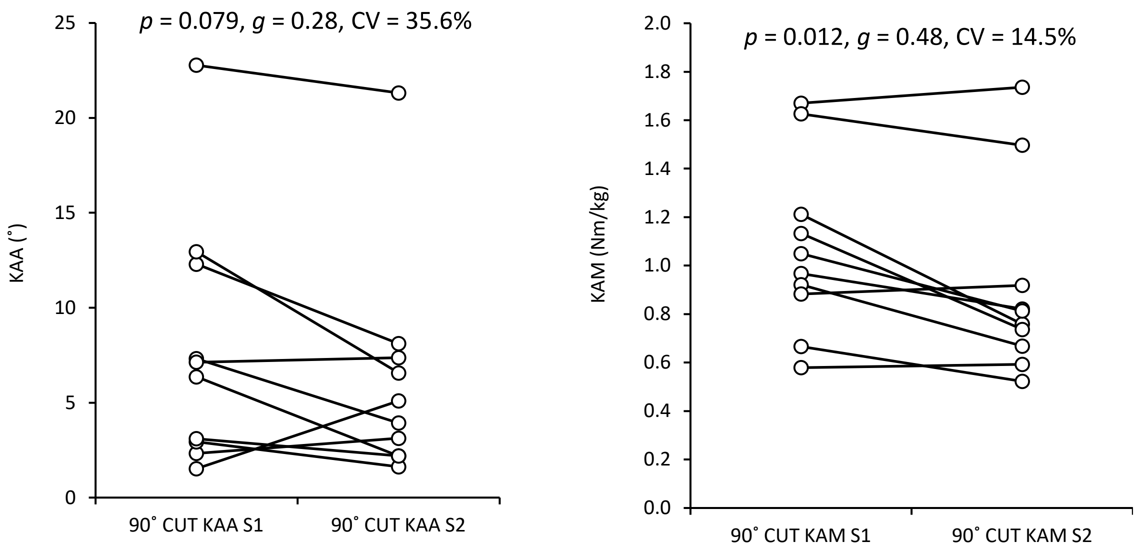


Figure 4.3. Differences in frontal plane parameters between sessions based on trial size 5 data (KAA: Knee abduction angle; KAM: Knee abduction moment; S: Session).

KAAs and KAMs have been identified as "high-risk" biomechanical deficits associated with increased non-contact ACL injury risk (183, 242, 298). As such, being able to reliably screen athletes who display these abnormal mechanics is important, so informed decisions can be made regarding the future training (241, 372, 436). Concerningly, peak KAAs and KAMs demonstrated high and unacceptable CV% (Tables 4.3 & 4.4, Appendix 3.1) within- and between-sessions, and also systematic bias between sessions was observed, with KAA and KAMs tending to reduce for session 2 and the ranking of subjects was also inconsistent (Figure 4.3). These results are similar to Sankey et al. (497) who observed high variability between sessions for frontal plane knee moments (24.1 Nm), while Mok et al. (370) observed a significant difference between sessions for KAMs during cutting ($p < 0.05$, $d = 0.17$), and reported only six of 12 subjects (top 30% of the sample) based on KAM were defined as "high-risk" for both sessions. Taken together, these findings indicate that different evaluations regarding an athlete's "injury risk profile" could be made between sessions, and the high CV% and systematic bias observed for KAAs and KAMs, reduces the certainty that any observed changes in these parameters are "real" or "meaningful". As such, because of the poor reliability observed for these frontal plane parameters, it is central that athletes are familiarised with the task to improve reliability measures, and it is recommended that future studies which monitor changes in these variables in response to training interventions acknowledge measurement error to establish "real" changes, and also include a CG to control for the effect familiarisation, in order to be confident that changes in COD injury risk parameters can be attributed to the intervention and not due to familiarisation.

Although the male soccer athletes in this study demonstrated low variability and high ICCs in the outcome measure (completion time) for cutting (i.e., endpoint variability), the technical execution of the cut (movement variability), as indicated by the SEM and CV% of sagittal plane joint angles, moments, and GRF, varied within- and between-sessions (Table 4.3, Appendix 3.1). Thus, these results indicate a range of slightly different movement patterns (i.e., differences in joint angles, moments, GRF production) were used by male soccer players to complete the cutting task. This finding is in line with previous studies that have shown movement variability in athletic tasks including basketball throwing, javelin throwing, baseball pitching, locomotion, golf swings, and triple jumping (21, 22, 59, 175, 466, 565, 609). Due to the task complexity of cutting and the multiple degrees of freedom often involved in the coordination and control of human movements, the fact the movement variability is present during COD is unsurprising, because complex interactions of multiple body components (muscles, joints and segments) are involved during task execution (22, 36, 219, 226, 466, 570). Movement variability is suggested to be neither good or bad, and is essentially part of the signal (internal source of variability) that cannot be eliminated (466). As such, every effort should be made to standardise and eliminate external sources of error such as inconsistent marker placement, data collection and collaboration issues, motivation, and fatigue.

4.6 LIMITATIONS

It should be noted that there are several limitations within this study. Firstly, the current study examined discrete parameters (i.e., peak values) and did not qualitatively compare the whole waveforms unlike Mok et al. (370). Alternatively, one-dimensional statistical parametric mapping

(SPM) could have also been performed to explore temporal differences for variables between trial sizes and sessions (497, 574). Secondly, the present study only examined lower-limb joint angle, moment, GRF, and GCT data; however, other variables such as joint powers, and trunk variables are also of interest. Thirdly, it is worth acknowledging that the 90° cutting task from a 15-m approach was a simple, repetitive, and prescriptive task which arguably may lack ecological validity to the CODs performed in multidirectional sport. Additionally, the study only examined two sessions, and may have benefited from a third session to determine if there was a stabilisation in outcome values and reliability. Finally, the findings from this study are limited and representative of the specific laboratory using the same data collection and analysis procedures, and the present study only examined a male soccer population. Thus, the generalisation of these results to different populations and laboratories should be made with caution. Consequently, future research should considerer investigating between-session COD biomechanics through temporal analysis, such as SPM, and examine coordination variability between-sessions, and different athletic populations.

4.7 CONCLUSION

In conclusion, PFC and FFC sagittal plane joint angles, GRFs, PFC GCTs, and KFMs were highly reliable within- and between- sessions during cutting across all trial sizes. Conversely, PFC and FFC ADFMs and HFMs displayed high levels of variability within- and between-session across all trial sizes. Specifically, injury risk parameters KIRMs, KAAs and KAMs displayed high and unacceptable levels of variability within- and between-sessions, tended to decrease for session 2, while participant ranking was inconsistent between sessions. As such, caution is advised when using injury risk parameters for diagnostic purposes because different clinical evaluations could be made between sessions. Furthermore, the high levels of variability observed for KIRMs, KAAs, and KAMs must also be acknowledged when longitudinally monitoring changes in performance to identify "real" changes, and highlight that a CG is imperative when monitoring changes in these parameters for greater certainty that "real" changes are observed.

In general, no substantial and minimal differences in reliability measures and outcome values for cutting biomechanical variables were revealed between trial sizes for within- and between-session data. As such, trial sizes of three to five (good trials that meet acceptable criteria) appear more than adequate to capture reliable cutting biomechanical data, with no additional benefit or apparent requirement for greater trial sizes. With this in mind, using lower trial sizes of three to five may reduce the time required for data collection and analysis for one subject, reduce financial costs due to less time for testing and analysis, and reduce the fatigue and potential injury risk for participants during maximal effort testing.

CHAPTER 5: Biomechanical Determinants of Performance and Injury Risk During Cutting: A "Performance-Injury Conflict"?

5.1 INTRODUCTION

An athlete's ability to change direction is one of the most important physical qualities for successful performance in multidirectional sports (51, 171, 283, 409, 487, 510, 549, 628), and is considered to provide the mechanical foundation for efficacious agility performance (409-412, 628). COD manoeuvres (side-steps, XOCs, split-steps, and pivots) are frequently performed in sports, such as soccer (51, 487), netball (184, 549), and rugby (499, 601, 632), with soccer players performing ~600 cuts of 0-90° (51) during match play, while directional changes of 45° and 90° have recently been established as frequently performed actions in netball (549). Specifically, side-step cutting actions are the most commonly performed attacking agility actions in netball (184), and are typically performed to create separation from an opponent to get into space and receive a pass. Moreover, side-steps are successful evasive manoeuvres in rugby and are linked to positive outcomes such as penetrating the defensive line (402, 601, 632). As such, developing an athlete's side-step cutting ability can be considered an important attribute to develop for successful performance in multidirectional sports.

Changing direction, particularly side-step manoeuvres, has been identified as a key action associated with non-contact ACL injuries in numerous multidirectional sports (soccer, rugby, handball, netball, Australian rules football, American football, and badminton) (52, 63, 94, 170, 270, 291, 298, 376, 426, 583), due to the potential to generate high multiplanar knee joint loading (flexion, rotation, and abduction moments) during the plant foot contact (39, 126, 127, 273, 301), thus increasing ACL strain (26, 288, 342, 424, 513). ACL injuries are debilitating and potentially career threatening, with short- and long-term consequences (financial, health, and psychological) (112, 241, 315, 333, 468). Specifically, an elevated and earlier risk of developing osteoarthritis is a primary concern associated with ACL injury (333, 571). An estimated two million ACL injuries occur worldwide (483), most of which typically require surgery (210); thus, extensive rehabilitation periods are required, resulting in prolonged absence from sport and the potential to lose sporting scholarships or contracts (375). While athletes who do successfully return to sport post ACL reconstruction, may demonstrate reduced sports-related performance, reduced number of appearances, and shorter career longevity (313, 368). Therefore, understanding the mechanics and techniques that can reduce the relative risk of injury during COD actions, while improving performance, are of great interest to researchers and practitioners working with multidirectional athletes.

Despite the importance of directional changes for sports performance and its association with ACL injury risk, it is somewhat surprising that the majority of studies into COD biomechanics investigate performance (143, 275, 338, 346, 498, 530, 531, 593, 596, 597) and injury risk determinants (117, 127, 139, 267, 272-274, 301, 358, 516, 519, 625) independently. From a performance perspective, greater braking and propulsive forces and impulses over short GCTs have been identified as mechanical determinants of faster COD speed performance (143, 228, 338, 346, 498, 530, 531, 596, 597). Additionally, whole-body kinetics and kinematics such as greater ankle power, ankle plantar-flexor moments, hip power and extensor moments, rapid knee and hip extension, wide lateral foot plants, torso lean and rotation, and low COM, are also associated with faster cutting performance (228, 346, 597); highlighting the importance of the lower-limb triple extensor musculature and trunk lean towards the intended direction of travel. Conversely, from an injury risk perspective, COD techniques with a wide lateral foot plant (127, 228, 272, 301), greater hip abduction angles (519, 593), increased IFPAs (274, 519), increased initial hip internal rotation angles (228, 358, 516, 519), greater peak and initial KAA (272, 274, 301, 358, 516), greater lateral trunk flexion (127, 141, 189, 267, 272), smaller KFAs (76, 593), and greater GRFs (273, 516, 519) are associated with greater peak KAMs and thus greater ACL strain (242, 342, 343, 512, 614). However, less is known regarding the mechanics and techniques necessary for optimal COD performance and how they relate and interact with injury risk (183, 228).

There is preliminary evidence, although limited, which indicates the techniques and mechanics required for faster COD performance are in direct conflict with the techniques and mechanics required for safer COD (i.e., lower knee joint loads) (115, 183, 228). For instance, COD techniques such as increased IFPAs and pelvic and hip internal rotation angles are associated with greater KAMs (127, 274, 519), but may be optimal for COD performance due to effective realignment of the whole-body COM into the new intended direction (229, 274). Extended knee postures (i.e., small KFAs) increase anterior tibial shear and subsequently strains the ACL (42, 43, 343, 344, 506, 615), yet increasing KFAs during side-stepping increases GCT and results in reduced exit velocity (115), thus negatively affecting performance. Greater KFMs (228) and posterior GRF (143, 202, 275) are associated with faster COD performance, but can also increase proximal anterior tibial shear (506, 631) and potential ACL loading (42, 43, 343, 615). Lateral trunk flexion has been shown to increase knee joint loading (126, 127, 272); however, this strategy may be adopted by athletes to deceive (feint) opponents (61, 62, 252). Importantly, wide lateral foot plants (126, 127, 228, 272, 301) are also associated with greater KAMs, where larger moment arms and KAMs are created with a more medial whole-body position with respect to the foot and COP positioning more lateral to the COM of the body and tibia (228, 274). However, a wide lateral foot plant is required for MLGRF and impulse generation to accelerate into the new direction (228, 261, 272, 597).

To the best of the author's knowledge, Havens and Sigward (228) is the first and only study to investigate the biomechanical determinants of cutting performance and injury risk, confirming that techniques required for faster performance are in direct conflict with reduced knee joint loading. The authors revealed faster cutting performance was associated with greater lateral foot plant distances, ML impulse, and internal hip rotation angles, though it is worth noting that greater KAMs were also observed with wider lateral foot plants, which may increase ACL injury risk. The study by Havens and Sigward (228) indicates that techniques required for faster cutting performance concurrently elevate KAMs, and a recent review by Fox (183) has highlighted that reducing "high-risk" postures (such as wide foot plants, lateral trunk flexion, increasing knee flexion, internal hip and foot progression angle) are viable strategies to reduce such knee joint loads, but could be to the detriment of faster performance. As athletes are driven by performance, they may be unlikely to adopt movement strategies which decrease knee injury risk if they do not result in effective performance (183, 228). Taken together, these studies suggest that there is a "performance-injury conflict" during COD, which is problematic for practitioners who aim to improve their athletes' performance and reduce injury risk. As such, further insight is required to improve our understanding of mechanics required for faster and safer COD.

Although providing further insight into the performance and injury risk determinants during cutting, Havens and Sigward (228) did not examine KIRMs. This absence is important because ACL strain is amplified when a combination of high frontal and transverse knee moments are generated in comparison to uniplanar loading (26, 288, 342, 424, 513). Additionally, Havens and Sigward (228) did not examine approach or exit velocities during the COD. Faster approach velocities and minimising velocity loss during cutting has been identified as a key determinant of faster performance (213, 275), and faster approach velocities concurrently elevate knee joint loading (i.e., KAMs, KFMs) (114, 148, 290, 403, 574). Moreover, cutting is a multistep action (11, 147, 273); however, Havens and Sigward (228) only investigated the FFC. Emerging research has demonstrated that greater braking forces displayed during the PFC (i.e., PFC dominant braking) is associated with faster COD performance (143, 147, 202, 275) and reduced KAMs in the FFC (272-274). Finally, there are only a limited number of studies that have comprehensively examined the whole-body biomechanical determinants of COD performance (228, 346, 530, 597) using 3D motion analysis, but these studies are low in sample size (n = 15-25). Therefore, the aim of this study was to expand on the work of Havens and Sigward (228) by investigating the biomechanical determinants of performance and injury risk during cutting, with a larger sample size, while examining the velocity profile (approach and exit), KIRMs, and PFC braking characteristics. Conducting such research into the relationship and interaction between performance and injury risk determinants during COD, may assist in the development of more effective ACL injury mitigation, COD speed, and agility programmes (183). It was hypothesised that the mechanical properties responsible for faster performance would concurrently increase knee joint loading.

5.2 METHODS

5.2.1 Research design

This study used a mixed, cross-sectional design to determine the relationship between COD biomechanics and COD performance (completion time, GCT, exit velocity) and injury risk (peak KAMs and peak KIRMs), following an associative strategy (17). In addition, a between-subject, comparative design was used to explore differences in COD biomechanics faster and slower and "higher-"and "lower-risk" (KAM) athletes. Participants performed six 90° cuts from a short (5-m) and long approach (15-m) for insight into low and high entry-velocity COD ability (410, 411) (Figure 3.1). 3D motion and GRF analysis was used to explore the joint kinetic, kinematic, and GRF determinants of performance and injury risk during cutting, similar to the procedures of previous research (228).

5.2.2 Participants

A minimum sample size of 48 participants was determined from an *a priori* power analysis using G*Power (Version 3.1, University of Dusseldorf, Germany) (172). This was based upon a previously reported correlation value of 0.472 (lateral foot plant distance to peak KAM) (228), a power of 0.95, and type 1 error or alpha level 0.05. As such, 61 male athletes (mean ± SD; age: 20.7 ± 3.8 years, height: 1.77 ± 0.06 m, mass: 74.7 ± 10.0 kg) from multiple sports (soccer n = 43, rugby n = 10, cricket n = 7, field hockey n = 1) participated in this study. For inclusion in the study, all athletes had played their respective sport for a minimum of five years and regularly performed one game and two structured skill-based sessions per week. All athletes were free from injury during the study and none of the athletes had suffered a prior traumatic knee injury such as an ACL injury. At the time of testing, players were currently in-season (competition phase). The investigation was approved by the Institutional Ethics Review Board (HSR1617-02), and all participants were informed of the benefits and risks of the investigation prior to signing institutionally approved consent or parental assent documents to participate in the study.

5.2.3 Procedures

For a detailed description of the warm up, short 90° cut, long 90˚ cut, marker placement, and 3D motion analysis procedures, please refer to Chapter 3.2 and 3.3 in the research methodology. Based on the results from Chapter 4 that 3 to 5 trials were considered sufficient to attain reliable cutting data, participants in the present study performed six trials of the short and long 90° cut (Figure 3.1) to obtain reliable data and accommodate for any errors that can only be noticed during data analysis. Marker and force data were collected over the PFC and FFC, and completion time was also measured for each trial. Completion time was measured using two sets of Brower timing lights (Draper, UT, USA) placed at the start and finish line.

5.2.4 Kinetic and kinematic Variables

Figures 2.7 and 2.8 outline deterministic models for greater peak KAMs and faster cutting performance based on the literature reviews in Chapter 2.3.3 and 2.3.4. As such, these variables form the cornerstone of variables of interest, and were subsequently investigated in the present study to better understand the potential "performance-injury conflict". A full description of variables along with definitions, abbreviations, and calculations are provided in Table 3.1 and 3.2. In summary, lower-limb joint moments were calculated over the FFC and lower-limb joint and trunk angles were also calculated and assessed at IC, peak, and ROM of the FFC. Peak and mean GRF braking and propulsive characteristics were also calculated as described in Table 3.1. PFC braking forces were also assessed for an indication of braking strategies, and horizontal COM velocity profiles at PFC IC to determine approach velocity, FFC touch-down, and at toe-off of the FFC to determine exit velocity, while cutting angle was also calculated (Tables 3.1 & 3.2). COD performance dependent variables were completion time, FFC GCT, and exit velocity, while injury risk dependent variables were peak KAMs and peak KIRMs and were used as surrogates of ACL injury risk.

5.3 STATISTICAL ANALYSES

All statistical analyses were performed in SPSS v 25 (SPSS Inc., Chicago, IL, USA) and Microsoft Excel (version 2016, Microsoft Corp., Redmond, WA, USA). Normality was inspected for all variables using a Shapiro-Wilks test. To explore the biomechanical determinants of performance and injury risk dependent variables, Pearson's (for parametric data) and Spearman's (for non-parametric data) correlations were used, similar to previous research (228). Correlations were evaluated as follows: trivial (0.00 - 0.09), small (0.10 – 0.29), moderate (0.30 – 0.49), large (0.50 – 0.69), very large (0.70 – 0.89), nearly perfect (0.90 – 0.99), and perfect (1.00) (255), with 95% CI also calculated. A correlation cut-off value of ≥ 0.40 was considered relevant according to Welch et al. (596) who also investigated the biomechanical determinants of cutting performance. Thus, correlations greater than this value are only reported in the subsequent tables. Stepwise multiple regression analysis was also performed to explore the relationship between the abovementioned variables and key primary performance and injury risk variables. Only significantly corelated variables that were parametric were considered for the Stepwise multiple regression analysis, and no more than 6 variables were inputted into the model to ensure a minimum 10:1 participant to independent variable ratio was present (578).

Comparisons in COD biomechanics between the faster and slower (top 33%, $n = 20$ vs. bottom 33%, $n = 20$ completion times/exit velocity), and "higher-" and "lower-risk" (top 33%, $n = 20$ vs. bottom 33%, $n = 20$ peak KAMs) cutting athletes were also performed using independent sample t-tests (parametric) or Mann-Whitney U tests (non-parametric), similar to previous research (143, 531). To explore the magnitude of differences between groups, Hedges' g effect sizes with 95% CIs were calculated as described previously (231) (Chapter 4.4), and interpreted as trivial (≤ 0.19), small (0.20 – 0.59), moderate (0.60 – 1.19), large (1.20 – 1.99), very large (2.00 – 3.99), and extremely large (≥ 4.00) (255). Statistical significance was defined as $p \leq 0.05$ for all tests. A minimum of four trials was used for each participant (274), and an average of individual trial peaks for each variable was used as previously recommended (116, 142).

5.4 RESULTS

Pearson's and Spearman's correlation values between COD biomechanical variables and cut completion times, GCT, exit velocity, peak KAM, and peak KIRM are presented in Tables 5.1 and 5.2 for short and long cut, respectively. The correlations presented in Tables 5.1 and 5.2 are presented in hierarchy order in terms of magnitude. The full correlation matrix is presented in Appendix 4.1.

5.4.1 Short cut biomechanical determinants

Completion time: short cut

Shorter completion times were largely associated with greater exit velocities and greater mean MLPFs; and moderately associated with greater peak MLPFs, greater peak RPFs, greater PFC mean HBFs, and greater approach velocities (Table 5.1). Completion time was non-parametric, thus Stepwise regression analysis could not be performed.

FFC GCT: short cut

Shorter GCTs were very largely associated with smaller lateral trunk flexion ROMs; largely associated with lower peak KFAs and ROM, lower peak HFAs and ROM, greater FFC touch down-velocities, greater exit velocities, greater lateral foot plant distances, and lower peak lateral trunk flexion angles; and moderately associated with greater peak KFMs and KIRMs (Table 5.1). FFC GCT was non-parametric, thus Stepwise regression analysis could not be performed.

Exit velocity: short cut

Faster exit velocities were very largely associated with greater mean MLPFs; largely associated with greater peak MLPFs, shorter completion times, greater peak and mean RPFs, greater FFC touch-down velocities, greater peak KFMs, shorter FFC GCTs, and greater mean and peak VPFs; and moderately associated with greater mean and peak HPFs, greater approach velocities, greater mean FFC RBFs, lower HFA ROM, and shorter PFC GCTs (Table 5.1). Stepwise multiple regression analysis revealed that FFC mean MLPF, FFC touch-down velocity, and peak KFM together could explain 64% (r = 0.800, adjusted 62%, p = 0.035) of the variation in exit velocity.

peak KAMs: short cut

Greater peak KAMs were largely associated with greater FFC touch-down velocities, lower HFA ROM, greater peak KIRMs; and moderately associated with greater peak KAAs (Table 5.1). Peak KAM was non-parametric, thus Stepwise regression analysis could not be performed.

peak KIRMs: short cut

Greater peak KIRMs were largely associated with FFC touch-down velocities, greater FFC mean HBFs and RBFs; and moderately associated with shorter FFC GCTs, greater peak FFC RBFs, lower HFA ROM, greater approach velocities, greater mean and peak VBFs, and greater peak HFMs. Peak KIRM was non-parametric, thus Stepwise regression analysis could not be performed.

Table 5.1. Short cut biomechanical variables associated with performance and injury risk variables presented in hierarchal order in terms of magnitude

Variable	↓ completion time $\rho \pm$ 95% CI
↑ exit velocity	0.621 ± 0.159, $p < 0.001$
↑ mean MLPF	0.609 ± 0.162, $p < 0.001$
↑ pk MLPF	-0.499 ± 0.192, $p < 0.001$
↑ PFC mean HBF	0.449 ± 0.204, $p < 0.001$
↑ mean RPF	-0.424 ± 0.209, $p = 0.001$
↑ approach velocity	-0.413 ± 0.211, $p = 0.001$

Variable	↓ FFC GCT r or $\rho \pm$ 95% CI
↓ lateral trunk flexion ROM	$r = 0.711 \pm 0.121$, $p < 0.001$
↓ KFA ROM	$r = 0.684 \pm 0.138$, $p < 0.001$
↓ pk KFA	$r = 0.621 \pm 0.159$, $p < 0.001$
↓ HFA ROM	$r = 0.654 \pm 0.148$, $p < 0.001$
↓ pk HFA	$r = 0.554 \pm 0.178$, $p < 0.001$
↑ FFC touch-down velocity	$r = 0.597 \pm 0.166$, $p < 0.001$
↑ exit velocity	$r = 0.547 \pm 0.180$, $p < 0.001$
↑ lateral foot plant distance	$r = 0.531 \pm 0.184$, $p < 0.001$
↓ pk lateral trunk flexion angle	$r = -0.531 \pm 0.184$, $p < 0.001$
↑ pk KFM	$r = -0.495 \pm 0.193$, $p < 0.001$
↑ pk KIRM	$\rho = -0.495 \pm 0.193$, $p < 0.001$

Variable	↑ exit velocity r or $\rho \pm$ 95% CI
↑ mean MLPF	$r = 0.726 \pm 0.123$, $p < 0.001$
↑ pk MLPF	$r = 0.687 \pm 0.137$, $p < 0.001$
↓ completion time	$\rho = -0.621 \pm 0.159$, $p < 0.001$
↑ mean RPF	$r = 0.611 \pm 0.162$, $p < 0.001$
↑ pk RPF	$\rho = -0.585 \pm 0.169$, $p < 0.001$
↑ FFC touch-down velocity	$r = 0.570 \pm 0.174$, $p < 0.001$
↑ pk KFM	$r = 0.557 \pm 0.177$, $p < 0.001$
↓ FFC GCT	$r = -0.547 \pm 0.180$, $p < 0.001$
↑ pk VPF	$r = 0.538 \pm 0.182$, $p < 0.001$
↑ mean VPF	$\rho = 0.512 \pm 0.189$, $p < 0.001$
↑ pk HPF	$r = -0.488 \pm 0.195$, $p < 0.001$
↑ mean HPF	$\rho = -0.462 \pm 0.201$, $p < 0.001$
↑ approach velocity	$r = 0.456 \pm 0.202$, $p < 0.001$
↑ mean FFC RBF	$r = 0.436 \pm 0.206$, $p < 0.001$
↓ HFA ROM	$\rho = -0.431 \pm 0.207$, $p = 0.001$
↓ PFC GCT	$r = -0.403 \pm 0.213$, $p = 0.001$

Variable	↑ pk KAM $\rho \pm$ 95% CI
↑ FFC touch-down velocity	0.527 ± 0.185, $p < 0.001$
↓ HFA ROM	-0.526 ± 0.185, $p < 0.001$
↑ pk KIRM	-0.522 ± 0.186, $p < 0.001$
↑ pk KAA	-0.455 ± 0.202, $p < 0.001$

Variable	↑ pk KIRMs $\rho \pm$ 95% CI
↑ FFC touch-down velocity	-0.588 ± 0.168, $p < 0.001$
↑ FFC mean HBF	-0.507 ± 0.190, $p < 0.001$
↑ FFC mean RBF	-0.501 ± 0.192, $p < 0.001$
↓ FFC GCT	0.495 ± 0.193, $p < 0.007$
↑ FFC pk RBF	-0.495 ± 0.193, $p < 0.001$
↓ HFA ROM	0.490 ± 0.194, $p < 0.001$
↑ approach velocity	-0.476 ± 0.198, $p < 0.001$
↑ FFC mean VBF	-0.468 ± 0.199, $p < 0.001$
↑ FFC pk VBF	-0.450 ± 0.203, $p < 0.001$
↑ pk HFM	0.411 ± 0.212, $p = 0.001$

Key: pk: peak; GCT: Ground contact time; PFC: Penultimate foot contact; FFC: Final foot contact; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFM: Ankle dorsi-flexion moment; HFA: Hip flexion angle; KFA: Knee flexion angle; ADFA: Ankle dorsi-flexion angle; RPF: Resultant propulsive force; RBF: Resultant braking force; VPF: Vertical propulsive force: VBF: Vertical braking force; HPF: Horizontal propulsive force; HBF: Horizontal braking force; MLPF: Medio-lateral propulsive force; ROM: Range of motion; IC: Initial contact; CI: Confidence interval; ↑: Greater/longer; ↓: lower/shorter

Moderate correlation (0.30-0.49) | **Large correlation (0.50-0.69)** | **Very large correlation (0.70-0.89)**

Table 5.2. Long cut biomechanical variables associated with performance and injury risk variables presented in hierarchal order in terms of magnitude

Variable	↓ completion time r or ρ ± 95% CI
↑ FFC touch-down velocity	ρ = -0.752 ± 0.113, p < 0.001
↑ exit velocity	r = -0.733 ± 0.121, p < 0.001
↑ approach velocity	ρ = -0.660 ± 0.146, p <0.001
↑ pk RPF	r = -0.641 ± 0.151, p < 0.001
↑ mean RPF	r = -0.530 ± 0.184, p < 0.001
↑ mean MLPF	r = -0.627 ± 0.157, p < 0.001
↑ pk MLPF	r = -0.588 ± 0.168, p < 0.001
↓ approach time	ρ = 0.620 ± 0.159, p < 0.001
↑ FFC mean HPF	r = 0.608 ± 0.163, p < 0.001
↓ FFC GCT	ρ = 0.581 ± 0.170, p < 0.001
↓ PFC GCT	ρ = 0.551 ± 0.179, p < 0.001
↑ PFC mean HBF	r = 0.551 ± 0.179, p < 0.001
↑ pk KIRM	ρ = -0.539 ± 0.182, p < 0.001
↑ FFC mean HBF	r = 0.535 ± 0.183, p < 0.001
↑ mean RBF	r = -0.484 ± 0.196, p < 0.001
↑ pk HPF	r = -0.460 ± 0.201, p < 0.001
↑ pk VPF	r = -0.449 ± 0.204, p < 0.001
↑ pk KAM	r = -0.412 ± 0.211, p = 0.001
↑ IFPA	r = -0.411 ± 0.212, p = 0.001
↓ HFA ROM	r = 0.406 ± 0.213, p = 0.001

Variable	↓ FFC GCT ρ ± 95% CI
↑ lateral foot plant distance	0.626 ± 0.157, p < 0.001
↓ KFA ROM	0.670 ± 0.143, p < 0.001
↓ pk KFA	0.616 ± 0.160, p < 0.001
↓ HFA ROM	0.658 ± 0.147, p < 0.001
↓ pk HFA	0.603 ± 0.164, p < 0.001
↓ lateral trunk flexion angle ROM	0.623 ± 0.158, p < 0.001
↓ lateral trunk flexion angle pk	-0.595 ± 0.166, p < 0.001

Variable	↑ exit velocity r or ρ ± 95% CI
↓ completion time	r = -0.733 ± 0.121, p < 0.001
↑ FFC touch-down velocity	ρ = 0.725 ± 0.124, p < 0.001
↑ pk MLPF	r = 0.651 ± 0.149, p < 0.001
↑ mean MLPF	r = 0.638 ± 0.153, p < 0.001
↓ FFC GCT	ρ = -0.569 ± 0.174, p < 0.001
↑ mean RPF	r = 0.568 ± 0.174, p < 0.001
↑ pk RPF	r = 0.549 ± 0.179, p < 0.001
↑ FFC pk VPF	r = 0.540 ± 0.182, p < 0.001
↑ approach velocity	ρ = 0.533 ± 0.184, p < 0.001
↑ pk HPF	r = 0.500 ± 0.192, p < 0.001
↑ mean VPF	r = 0.499 ± 0.192, p < 0.001
↓ PFC GCT	ρ = -0.484 ± 0.196, p < 0.001
↑ KFM	r = 0.482 ± 0.196, p = 0.001
↓ HFA ROM	ρ = 0.470 ± 0.199, p = 0.001
↑ FFC mean VBF	r = 0.456 ± 0.202, p = 0.001
↑ PFC mean HBF	r = -0.430 ± 0.208, p = 0.001

Variable	↑ pk KAM r or ρ ± 95% CI
↑ pk KIRM	ρ = -0.557 ± 0.172, p < 0.001
↑ FFC touch-down velocity	ρ = -0.491 ± 0.194, p = 0.001
↑ pk KAA	r = -0.468 ± 0.199, p < 0.001
↑ FFC mean VBF	r = 0.497 ± 0.193, p < 0.001
↑ FFC mean HBF	r = 0.434 ± 0.207, p < 0.001
↑ FFC mean RBF	r = 0.488 ± 0.195, p < 0.001
↓ completion time	r = -0.412 ± 0.211, p = 0.001

Variable	↑ pk KIRM ρ ± 95% CI
↑ pk KAMs	-0.557 ± 0.172, p < 0.001
↑ FFC touch-down velocity	-0.551 ± 0.179, p < 0.001
↓ completion time	0.539 ± 0.182, p < 0.001
↑ approach velocity	-0.534 ± 0.182, p < 0.001
↑ FFC pk RBF	-0.505 ± 0.191, p < 0.001
↑ FFC pk VBF	-0.475 ± 0.198, p < 0.001
↑ FFC mean VBF	-0.468 ± 0.199, p < 0.001
↑ FFC pk RBF	-0.458 ± 0.202, p < 0.001

Key: pk: peak; GCT: Ground contact time; PFC: Penultimate foot contact; FFC: Final foot contact; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFM: Ankle dorsi-flexion moment; HFA: Hip flexion angle; KFA: Knee flexion angle; ADFA: Ankle dorsi-flexion angle; RPF: Resultant propulsive force; RBF: Resultant braking force; VPF: Vertical propulsive force: VBF: Vertical braking force; HPF: Horizontal propulsive force; HBF: Horizontal braking force; MLPF: Medio-lateral propulsive force; ROM: Range of motion; IC: Initial contact; IFPA: Initial foot progression angle; ↑: Greater/longer; ↓: lower/shorter

Moderate correlation (0.30-0.49)	Large correlation (0.50-0.69)	Very large correlation (0.70-0.89)

5.4.2 Long cut determinants

Completion times: long cut

Shorter completion times were very largely associated with greater FFC touch-down and exit velocities; largely associated with faster approach velocities, greater peak and mean RPFs and MLPFs, shorter approach times, greater mean HPFs, greater peak KIRMs, shorter PFC and FFC GCTs, and greater PFC and FFC mean HBFs; and moderately associated with greater mean FFC RBFs, greater peak VPFs and HPFs, greater peak KAMs, greater IFPAs, and lower HFA ROM (Table 5.2). Stepwise multiple regression analysis revealed that exit velocity, peak RPF, PFC mean HBF, and IFPA together could explain 64% (r = 0.801, adjusted 62%, p = 0.048) of the variation in completion time.

FFC GCT: long cut

Shorter FFC GCTs were largely associated with wider lateral foot plant distances, lower peak KFAs and ROM, lower peak HFAs and ROM, and lower peak lateral trunk flexion angles and ROM (Table 5.2). FFC GCT was non-parametric, thus Stepwise regression analysis could not be performed.

Exit velocity: long cut

Faster exit velocities were very largely associated with shorter completion times and greater FFC touch-down velocities; largely associated with greater mean and peak MLPFs and RPFs, shorter FFC GCTs, greater peak VPFs and HPFs, and greater approach velocities; and moderately associated with greater mean VPFs, shorter PFC GCTs, greater peak KFMs, lower HFA ROM, greater FFC mean VBFs, and greater PFC mean HBFs (Table 5.2). Stepwise multiple regression analysis revealed that FFC peak MLPF and peak KFMs together could explain 48% (r = 0.690, adjusted 46%, p = 0.019) of the variation in exit velocity.

peak KAMs: long cut

Greater peak KAMs were largely associated with greater peak KIRMs; and moderately associated with greater FFC touch-down velocities, greater peak KAAs, greater FFC mean VBFs, HBFs, and RBFs, and shorter completion times (Table 5.2). Stepwise multiple regression analysis revealed that FFC mean VBF and peak KAA together could explain 43% (r = 0.652, adjusted 41%, p < 0.001) of the variation peak KAM.

peak KIRM: long cut

Greater peak KIRMs were largely associated with greater peak KAMs, greater FFC touch-down velocities, shorter completion times, greater approach velocities, and greater FFC peak RBFs; and moderately associated with greater FFC mean and peak VBFs, and greater FFC peak RBFs (Table 5.2). Peak KIRM was non-parametric, thus Stepwise regression analysis could not be performed.

Normalised moments by approach velocity

To control for the effect of approach velocity, correlations between COD biomechanics and peak KAMs, and KIRMs normalised by approach velocity are presented in Appendix 4.2, and in general substantiate the current findings presented above.

5.4.3 Between-group comparisons

Comparisons in COD biomechanics between faster and slower athletes based on completion times and exit velocities, and athletes with higher and lower peak KAMs are presented in Tables 5.3 & 5.4, containing *p* values and effect sizes. Appendix 4.3 presents the full descriptive statistics, *p* values, and effect sizes with 95% CI for all variables and comparisons.

5.4.3.1 Short cut comparisons

Fast vs. Slow completion time short cut biomechanical comparisons

Very large significant differences were revealed between faster and slower athletes based on their completion times (Table 5.3). Faster athletes, compared to slower, produced significantly greater propulsive forces in shorter GCTs which led to greater exit velocities (moderate to large ES); significantly greater PFC mean HBFs, approach and FFC touch-down velocities (moderate ES); and significantly greater peak HFMs, and greater forward trunk inclination angles (moderate ES) (Table 5.3). Incidentally, peak KIRMs were significantly greater for faster athletes compared to slower athletes (small ES) (Table 5.3).

Fast vs. Slow exit velocity short cut biomechanical comparisons

Extremely large significant differences were revealed between faster and slower athletes based on their exit velocities (Table 5.3). Faster athletes, compared to slower, produced significantly greater propulsive forces in shorter GCTs which led to greater exit velocities and completion times (moderate to large ES); greater FFC mean VBFs, HBFs, and RBFs, greater approach and FFC touch-down velocities, and greater peak KFMs and ADFMs (moderate to large ES); and smaller peak HFAs and ROM, greater

FFC ADFA ROM, greater IFPAs, and lower peak lateral trunk flexion angles and ROM (moderate ES) (Table 5.3). Incidentally, peak KAMs, peak KIRMs, and peak KAAs were significantly greater for faster athletes compared to slower athletes (moderate ES) (Table 5.3).

Low vs. High KAM short cut biomechanical comparisons

Extremely large significant differences were revealed between low and high KAM athletes (Table 5.3). Low KAM athletes, compared to high, demonstrated significantly lower peak KIRMs and lower peak and initial KAAs (moderate to large ES); lower approach and FFC touch-down velocities, and greater reductions in velocity over the PFC (moderate to large ES); lower HBF ratios (i.e., greater PFC braking emphasis) and lower braking forces over longer FFC GCTs (small to moderate ES); lower peak KFMs, greater hip and KFA ROM, lower ADFA ROM, and narrower lateral foot plant distances (moderate ES) (Table 5.3). Incidentally, low KAM athletes displayed lower exit velocities and longer GCTs (moderate ES) (Table 5.3).

5.4.3.2 Long cut comparisons

Fast vs. Slow completion time long cut biomechanical comparisons

Extremely large significant differences were revealed between faster and slower athletes based on their completion times (Table 5.4). Faster athletes produced significantly greater propulsive forces (moderate to large ES) over shorter FFC GCTs (large ES) which led to greater exit velocities (very large ES); greater PFC peak and mean HBFs (moderate to large ES); greater approach and FFC touch-down velocities, and shorter approach times (large to very large ES); and greater peak ADFMs, lower peak HFA and ROM, increased pelvic rotation and IFPAs (moderate ES) (Table 5.4). Incidentally, peak KIRMs and KAMs for faster athletes were significantly greater than slower athletes (large and moderate ES) (Table 5.4).

Fast vs. Slow exit velocity long cut biomechanical comparisons

Extremely large significant differences were revealed between faster and slower athletes based on their exit velocities (Table 5.4). Faster athletes, compared to slower, produced significantly greater propulsive forces over shorter GCTs which led to greater exit velocities and completion times (moderate to very large ES); greater PFC mean HBFs and RBFs, greater FFC mean HBFs and RBFs (moderate to large ES); greater approach and FFC touch-down velocities, shorter approach times (large to very large ES); greater peak KFMs and ADFMs, lower peak HFAs and ROM, greater ADFA ROM (moderate to large ES); lower peak lateral trunk flexion angles and ROM, and greater IFPAs (moderate

ES) (Table 5.4). Incidentally, peak KAMs and peak KIRMs were significantly greater than slower athletes (moderate and large ES) (Table 5.4).

<u>Low vs. High KAM long cut biomechanical comparisons</u>

Extremely large significant differences were revealed between low and high KAM athletes (Table 5.4). Low KAM athletes, compared to high, demonstrated significantly lower peak KIRMs and lower peak and initial KAAs (moderate to large ES); lower approach and FFC touch-down velocities (moderate to large ES); lower FFC braking forces in longer FFC GCTs, and also displayed lower HBF ratios (small to moderate ES); lower peak KFMs, greater hip and KFA ROM, and narrower lateral foot plant distances (moderate ES) (Table 5.4). Incidentally, low KAM athletes displayed lower exit velocities, longer FFC GCTs, and longer completion times compared to high KAM athletes (moderate to large ES) (Table 5.4).

Table 5.3. Comparisons in short cut biomechanics between Fast vs. Slow and Low vs. High KAM athletes

	Variable	Completion time Fast vs. Slow	Variable	Exit velocity Fast vs. Slow	Variable	pk KAM Low vs. High
Performance	↓ completion time ↓ GCT ↑ exit velocity ↑ approach velocity ↑ FFC touch-down velocity	$p < 0.001$, $g = -2.55 \pm 0.83$ $p = 0.020$, $g = -0.75 \pm 0.64$ $p < 0.001$, $g = 1.54 \pm 0.71$ $p = 0.001$, $g = 1.12 \pm 0.67$ $p = 0.012$, $g = 0.81 \pm 0.64$	↓ completion time ↓ GCT ↑ exit velocity ↑ approach velocity ↑ FFC touch-down velocity	$p < 0.001$, $g = -0.96 \pm 0.63$ $p < 0.001$, $g = -1.25 \pm 0.63$ $p < 0.001$, $g = 4.32 \pm 1.13$ $p = 0.005$, $g = 0.94 \pm 0.64$ $p < 0.001$, $g = 1.28 \pm 0.67$	↑ GCT ↓ exit velocity ↓ approach velocity ↑ PFC velocity reduction ↓ FFC touch-down velocity	$p = 0.012$, $g = 0.82 \pm 0.68$ $p = 0.002$, $g = -1.02 \pm 0.66$ $p = 0.028$, $g = -0.71 \pm 0.65$ $p = 0.040$, $g = -0.60 \pm 0.62$ $p < 0.001$, $g = -1.21 \pm 0.68$
Injury risk	↑ pk KIRM	$p = 0.024$, $g = 0.57 \pm 0.63$	↑ pk KAM ↑ pk KIRM ↑ pk KAA	$p = 0.022$, $g = 0.74 \pm 0.64$ $p = 0.046$, $g = -0.65 \pm 0.68$ $p = 0.029$, $g = -0.70 \pm 0.64$	↓ pk KAM ↓ pk KIRM ↓ lower pk and IC KAA	$p < 0.001$, $g = -4.49 \pm 1.16$ $p < 0.001$, $g = 1.24 \pm 0.64$ $p = 0.002$, $g = 1.01 \pm 0.64$ $p = 0.044$, $g = 0.64 \pm 0.63$
Propulsive force	↑ pk VPF ↑ mean VPF ↑ pk HPF ↑ mean HPF ↑ pk MLPF ↑ mean MLPF ↑ pk RPF ↑ mean RPF	$p = 0.043$, $g = 0.46 \pm 0.63$ $p = 0.063$, $g = 0.69 \pm 0.64$ $p = 0.018$, $g = -0.70 \pm 0.64$ $p = 0.007$, $g = -0.84 \pm 0.65$ $p < 0.001$, $g = 1.05 \pm 0.66$ $p < 0.001$, $g = 1.71 \pm 0.72$ $p = 0.009$, $g = 0.64 \pm 0.64$ $p = 0.001$, $g = 1.06 \pm 0.66$	↑ pk VPF ↑ mean VPF ↑ pk HPF ↑ mean HPF ↑ pk MLPF ↑ mean MLPF ↑ pk RPF ↑ mean RPF	$p < 0.001$, $g = 1.21 \pm 0.63$ $p < 0.001$, $g = 1.30 \pm 0.62$ $p = 0.001$, $g = -1.11 \pm 0.64$ $p = 0.002$, $g = -0.96 \pm 0.63$ $p < 0.001$, $g = 1.83 \pm 0.63$ $p < 0.001$, $g = 1.97 \pm 0.63$ $p < 0.001$, $g = 1.38 \pm 0.63$ $p < 0.001$, $g = 1.58 \pm 0.63$	↑ pk HPF	$p = 0.036$, $g = 0.66 \pm 0.67$
Braking force	↑ PFC mean HBF	$p = 0.001$, $g = -1.08 \pm 0.66$	↑ FFC mean VBF ↑ FFC mean HBF ↑ FFC mean RBF	$p = 0.011$, $g = 0.70 \pm 0.64$ $p = 0.041$, $g = -0.66 \pm 0.65$ $p = 0.011$, $g = -0.83 \pm 0.65$	↓ FFC mean VBF ↓ FFC mean HBF ↓ FFC mean RBF ↓ pk RBF ↓ pk HBF ratio	$p = 0.016$, $g = -0.79 \pm 0.64$ $p = 0.010$, $g = 0.84 \pm 0.64$ $p = 0.010$, $g = -0.84 \pm 0.65$ $p = 0.031$, $g = -0.70 \pm 0.63$ $p = 0.049$, $g = -0.51 \pm 0.62$
Joint moment	↑ pk HFM	$p = 0.027$, $g = 0.71 \pm 0.64$	↑ pk KFM ↑ pk ADFM	$p < 0.001$, $g = 1.60 \pm 0.64$ $p = 0.039$, $g = -0.66 \pm 0.63$	↓ pk KFM	$p = 0.014$, $g = -0.80 \pm 0.71$
Sag joint angle	↓ HFA ROM	$p = 0.183$, $g = 0.55 \pm 0.63$	↓ pk HFA ↓ HFA ROM ↑ ADFA ROM	$p = 0.008$, $g = -0.86 \pm 0.64$ $p = 0.007$, $g = -1.06 \pm 0.67$ $p = 0.020$, $g = 0.73 \pm 0.64$	↑ HFA ROM ↑ KFA ROM ↓ ADFA ROM	$p < 0.001$, $g = 1.13 \pm 0.66$ $p = 0.040$, $g = 0.66 \pm 0.63$ $p = 0.042$, $g = -0.65 \pm 0.64$
Trunk	↑ FFC trunk inclination angles ↑ PFC trunk inclination angles	$p = 0.020$, $g = 0.65 \pm 0.64$ $p = 0.002$, $g = 0.79 \pm 0.64$	↓ pk lateral trunk flexion angle and ROM	$p = 0.017$, $g = 0.77 \pm 0.62$ $p = 0.014$, $g = -0.80 \pm 0.64$		
Hip, pelvis, foot			↑ IFPA	$p = 0.010$, $g = 0.84 \pm 0.63$	↓ lateral foot plant distance	$p = 0.032$, $g = 0.71 \pm 0.63$

Key: pk: peak; GCT: Ground contact time; PFC: Penultimate foot contact; FFC: Final foot contact; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFM: Ankle dorsi-flexion moment; HFA: Hip flexion angle; KFA: Knee flexion angle; ADFA: Ankle dorsi-flexion angle; RPF: Resultant propulsive force; RBF: Resultant braking force; VPF: Vertical propulsive force: VBF: Vertical braking force; HPF: Horizontal propulsive force; HBF: Horizontal braking force; MLPF: Medio-lateral propulsive force; ROM: Range of motion; IC: Initial contact; IFPA: Initial foot progression angle ↑: Greater/longer; ↓: lower/shorter. Italic denotes non-parametric.

Small ES (0.20-0.59)	Moderate ES (0.60-1.19)	Large ES (1.20-1.99)	Very Large ES (2.00-3.99)	Extremely Large ES (≥4.00)

Table 5.4. Comparisons in long cut biomechanics between Fast vs. Slow and Low vs. High KAM participants

	Variable	Completion time: Fast vs. Slow	Variable	Exit velocity: Fast vs. Slow	Variable	pk KAM: Low vs. High
Performance	↓ completion time	$p < 0.001, g = -4.80 \pm 1.22$	↓ completion time	$p < 0.001, g = -2.97 \pm 0.90$	↑ completion time	$p < 0.001, g = 1.22 \pm 0.67$
	↓ FFC GCT	$p < 0.001, g = 1.26 \pm 0.68$	↓ FFC GCT	$p = 0.001, g = -1.15 \pm 0.67$	↑ PFC GCT	$p = 0.003, g = 0.89 \pm 0.65$
	↓ PFC GCT	$p < 0.001, g = 1.18 \pm 0.67$	↓ PFC GCT	$p < 0.001, g = -1.45 \pm 0.70$	↑ FFC GCT	$p = 0.021, g = 0.74 \pm 0.64$
	↑ exit velocity	$p < 0.001, g = 1.54 \pm 0.71$	↑ exit velocity	$p < 0.001, g = 4.52 \pm 1.17$	↓ exit velocity	$p = 0.002, g = -1.05 \pm 0.66$
	↑ approach velocity	$p < 0.001, g = 2.06 \pm 0.77$	↑ approach velocity	$p < 0.001, g = 1.59 \pm 0.71$	↓ approach velocity	$p = 0.004, g = -0.94 \pm 0.65$
	↑ FFC touch-down velocity	$p < 0.001, g = 2.57 \pm 0.84$	↑ FFC touch-down velocity	$p < 0.001, g = 2.54 \pm 0.83$	↓ FFC touch-down velocity	$p < 0.001, g = -1.35 \pm 0.69$
	↓ approach times	$p < 0.001, g = -1.54 \pm 0.71$	↓ approach times	$p < 0.001, g = -1.26 \pm 0.68$		
Injury risk	↑ pk KAM	$p = 0.001, g = 1.01 \pm 0.66$	↑ pk KAM	$p = 0.006, g = 0.90 \pm 0.65$	↓ pk KAM	$p < 0.001, g = -4.60 \pm 1.18$
	↑ pk KIRM	$p < 0.001, g = -1.35 \pm 0.69$	↑ pk KIRM	$p < 0.001, g = -1.21 \pm 0.67$	↓ pk KIRM	$p < 0.001, g = 1.38 \pm 0.69$
					↓ pk KAA	$p < 0.001, g = 1.26 \pm 0.68$
					↓ IC KAA	$p = 0.023, g = 0.74 \pm 0.63$
Propulsive force	↑ pk VPF	$p = 0.007, g = 0.88 \pm 0.65$	↑ pk VPF	$p < 0.001, g = 1.36 \pm 0.69$	↑ pk HPF	$p = 0.036, g = 0.65 \pm 0.64$
	↑ mean VPF	$p = 0.001, g = 0.99 \pm 0.66$	↑ mean VPF	$p < 0.001, g = 1.41 \pm 0.69$		
	↑ pk HPF	$p < 0.001, g = -1.30 \pm 0.68$	↑ pk HPF	$p < 0.001, g = -1.37 \pm 0.69$		
	↑ mean HPF	$p < 0.001, g = -1.20 \pm 0.67$	↑ mean HPF	$p = 0.002, g = 1.02 \pm 0.66$		
	↑ pk MLPF	$p < 0.001, g = 1.50 \pm 0.70$	↑ pk MLPF	$p < 0.001, g = 1.82 \pm 0.74$		
	↑ mean MLPF	$p < 0.001, g = 1.82 \pm 0.74$	↑ mean MLPF	$p < 0.001, g = 2.19 \pm 0.78$		
	↑ pk RPF	$p = 0.001, g = 1.10 \pm 0.67$	↑ pk RPF	$p < 0.001, g = 1.50 \pm 0.70$		
	↑ mean RPF	$p < 0.001, g = 1.37 \pm 0.69$	↑ mean RPF	$p < 0.001, g = 1.75 \pm 0.73$		
Braking force	↑ PFC mean HBF	$p = 0.023, g = -0.74 \pm 0.64$	↑ PFC mean HBF	$p = 0.002, g = -1.01 \pm 0.66$	↓ FFC pk VBF	$p = 0.049, g = -0.49 \pm 0.63$
	↑ PFC pk HBF	$p < 0.001, g = -1.81 \pm 0.74$	↑ PFC mean RBF	$p = 0.043, g = 0.65 \pm 0.64$	↓ FFC mean HBF	$p = 0.005, g = 0.91 \pm 0.65$
	↑ PFC mean RBF	$p = 0.026, g = 0.72 \pm 0.64$	↑ FFC mean HBF	$p = 0.001, g = -1.08 \pm 0.66$	↓ FFC mean RBF	$p = 0.001, g = -1.11 \pm 0.65$
	↑ PFC pk RBF	$p = 0.014, g = 0.80 \pm 0.64$	↑ FFC mean RBF	$p < 0.001, g = 1.24 \pm 0.68$	↓ mean HBF ratio	$p = 0.127, g = -0.56 \pm 0.63$
	↑ FFC mean HBF	$p < 0.001, g = -1.25 \pm 0.68$				
Joint moment	↑ pk ADFM	$p = 0.038, g = -0.66 \pm 0.64$	↑ pk KFM	$p < 0.001, g = 1.26 \pm 0.68$	↓ pk KFM	$p = 0.023, g = -0.74 \pm 0.64$
			↑ pk ADFM	$p = 0.003, g = -0.99 \pm 0.66$		
Sag joint angle	↓ pk HFA and ROM	$p = 0.022, g = -0.74 \pm 0.64$	↓ pk HFA	$p = 0.002, g = -1.04 \pm 0.66$	↑ peak HFA	$p = 0.026, g = 0.72 \pm 0.64$
		$p = 0.003, g = -1.11 \pm 0.67$	↓ HFA ROM	$p < 0.001, g = -1.41 \pm 0.70$	↑ HFA ROM	$p = 0.012, g = 1.09 \pm 0.66$
			↑ ADFA ROM	$p = 0.055, g = 0.61 \pm 0.63$	↑ KFA ROM	$p = 0.057, g = 0.61 \pm 0.63$
Trunk			↓ pk lateral trunk flexion angle and ROM	$p = 0.032, g = 0.69 \pm 0.64$		
				$p = 0.015, g = -0.80 \pm 0.6$		
Hip, pelvis, foot	↑ IFPA	$p = 0.003, g = 0.97 \pm 0.66$	↑ IFPA	$p = 0.003, g = 0.97 \pm 0.63$	↓ lateral foot plant distance	$p = 0.027, g = 0.72 \pm 0.64$
	↑ pelvic rotation	$p = 0.037, g = 0.67 \pm 0.64$				

Key: pk: peak; GCT: Ground contact time; PFC: Penultimate foot contact; FFC: Final foot contact; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFM: Ankle dorsi-flexion moment; HFA: Hip flexion angle; KFA: Knee flexion angle; ADFA: Ankle dorsi-flexion angle; RPF: Resultant propulsive force; RBF: Resultant braking force; VPF: Vertical propulsive force: VBF: Vertical braking force; HPF: Horizontal propulsive force; HBF: Horizontal braking force; MLPF: Medio-lateral propulsive force; ROM: Range of motion; IC: Initial contact; IFPA: Initial foot progression angle; ↑: Greater/longer; ↓: lower/shorter. Italic denotes non-parametric.

Small ES (0.20-0.59)	Moderate ES (0.60-1.19)	Large ES (1.20-1.99)	Very Large ES (2.00-3.99)	Extremely Large ES (≥4.00)

5.5 DISCUSSION

The aim of this study was to expand on the work of Havens and Sigward (228) by investigating the whole-body biomechanical determinants of performance and injury risk during cutting, while also examining approach and exit velocity, KIRMs, and PFC braking characteristics. The results of this study substantiate Havens and Sigward (228) and the study hypothesis, whereby techniques and mechanics associated with faster performance (i.e., faster approach and exit velocity, greater FFC braking forces in short GCTs, greater KFMs, smaller FFC hip and KFAs and ROM, wider lateral foot plant distances, and greater IFPAs) are in direct conflict with safer COD mechanics (i.e., reduced knee joint loading) (Tables 5.1-5.4), and support the concept that a "performance-injury conflict" exists during cutting (115, 148, 183). The following discussion will firstly discuss the biomechanical determinants of performance and injury risk separately, before discussing the "performance-injury conflict" that is present during cutting.

5.5.1 Biomechanical determinants of performance

Focusing on cutting biomechanics solely from a performance perspective, correlational analysis and fast versus slow comparisons indicated that faster athletes demonstrated greater velocity at key instances over the PFC and FFC and greater peak and mean braking and propulsive forces, particularly MLPFs and RPFs, and these forces were produced over substantially shorter GCTs which led to superior exit velocities (Tables 5.1-5.4). These findings corroborate the results of previous research that have identified greater propulsive forces (143, 228, 530, 531, 596) and shorter GCTs (143, 338, 346, 498, 531, 596, 597) as determinants of faster COD performance. The fact that greater propulsive forces and shorter GCTs were determinants of faster performance are unsurprising because greater propulsive forces increase impulse which, based on Newton's 2nd law of motion and the impulse-momentum relationship, leads to greater changes in momentum, thus velocity (56, 634), and shorter GCTs indicate less time is spent performing the COD action. Additionally, greater propulsive forces applied in the ML direction results in greater net acceleration into the intended direction of travel (379). Moreover, due to the plyometric nature and reactive strength element required during the plant phase of cutting (205, 453, 596, 629), shorter GCTs are advantageous for faster performance because less time is required to generate braking and propulsive impulse and may facilitate greater utilisation of the stretch shortening cycle (SSC) (55, 596). Slower athletes, in contrast, may require a longer GCT to accept and transmit the forces, and therefore most likely display inferior reactive strength capabilities.

Greater approach velocities, velocities at key instances and velocity maintenance during the PFC and FFC were also associated with faster performance and were characteristics displayed by faster athletes (Tables 5.1-5.4). These results support previous research that found greater PFC approach velocities (275) during 180° turns, and greater minimum speed during 45-90˚ cutting (213) as determinants of faster COD performance. Greater velocities will enable athletes to cover greater distances in shorter times; thus, it is unsurprising that greater velocities were associated with faster completion times and exit velocities. Interestingly, faster athletes demonstrated greater approach, FFC touch-down, and exit velocities which highlights the importance of velocity maintenance and minimising the decline in velocity for faster cutting. Faster athletes also demonstrated greater FFC braking forces, which is suggested to be effective for reducing momentum and facilitating superior exit velocity due to increased storage and utilisation of elastic energy as the muscle lengthens under an eccentric load (258, 530, 531). Moreover, greater PFC HBFs were also associated with faster cutting performance and demonstrated by faster athletes (Tables 5.1-5.4). This finding substantiates the results observed in 180° turning that observed greater PFC HBFs were associated with faster COD performance (143, 147, 202, 275), while this is the first study, to the best of the author's knowledge, to confirm the importance of PFC HBFs for faster 90° cutting. Cutting is a multistep action (11, 273, 403, 445, 495), and displaying greater braking forces in a posteriorly directed direction facilitates reductions in momentum (net deceleration) to permit effective WA (147), thus rationalising the importance of PFC HBFs for faster cutting performance.

Greater peak HFMs and KFMs were demonstrated by faster athletes and associated with faster cutting performance (Tables 5.1-5.4). Havens and Sigward (228) also found peak KFMs were associated with faster cutting performance, while peak KFMs have been shown to increase linearly with approach velocity (114, 148, 290, 574) and posterior GRF (506, 631) (also associated with faster cutting performance). Nonetheless, although partly a product of (and influenced by) approach velocity (114, 148, 290, 574), cutting places high eccentric demands on the knee extensor musculature (37, 212, 341) and thus, KFMs are considered to play an important role during the braking phase of CODs. As such, Havens and Sigward (228) suggested that greater KFMs may permit a more rapid transition during the redirection phase of a COD (229), and the greater eccentric loading of the knee during the WA may lead to greater stimulation of the SSC (increases storage and release of elastic energy) leading to greater force production during push-off (55, 258, 317). Moreover, further down the lower-limb chain, faster athletes also demonstrated greater ADFMs (Tables 5.1-5.4), which substantiates previous research which found greater ADFMs and ankle power were associated with faster cutting performance (228, 346). This finding is unsurprising because concentric ankle power is pivotal for impulsive activities such as sprinting (123) and vertical jumping (572). Collectively, these results highlight the importance of the lower-limb triple-extension musculature and force generating capacity for cutting performance.

Technical and kinematic variables were also associated with faster cutting performance and key differences also existed between faster and slower athletes (Tables 5.1-5.4). For example, faster athletes displayed lower peak hip and KFAs and ROM, while displaying greater ADFA ROM (Tables 5.3-5.4). These aforementioned variables were also associated with shorter GCTs (Tables 5.1 & 5.2). Dai et al. (115) has previously shown that increasing KFA during COD negatively affects GCT duration and subsequent exit velocity. Additionally, Welch et al. (597) has recently identified resisting hip flexion and rapid hip and knee extension as key factors associated with faster cutting performance. It is postulated that the decreased FFC hip and knee flexion ROM demonstrated by the faster athletes in the present study, in combination with greater GRF, could indicate a "stiffer" strategy to permit a rapid transition from braking (WA) to drive-off (propulsion), thus facilitating a shorter GCT and greater utilisation of the SSC for more effective force generation (55, 317). Moreover, in terms of trunk control, faster cutting athletes demonstrated smaller peak lateral trunk flexion angles and ROM, while these aforementioned variables were also associated with shorter GCTs. Reducing lateral trunk flexion will have the effect of shifting the COM more towards the intended direction of travel (293, 346), which could be potentially attributed to superior pelvic stability (464). Additionally, faster athletes also displayed greater pelvic rotation and IFPAs, which may indicate more effective alignment of the whole-body COM towards the intended direction of travel (118, 228, 229). Consequently, these results substantiate previous research that has highlighted the importance of trunk and pelvic stability and control, while orientating the trunk and pelvis towards the intended direction of travel with faster COD performance (346, 498, 597).

Overall, these results indicate that velocity, mechanical, braking and propulsive forces, and technical factors are linked to faster cutting performance (Tables 5.1-5.4). Faster approach velocities and velocity maintenance over key instances of the PFC and FFC, greater braking and propulsive forces over shorter GCTs are strong determinants of faster performance and are therefore mechanical qualities which practitioners may look to develop in their athletes' physical preparation training programmes. Additionally, technical and joint kinematic factors such as greater ADFA ROM, and smaller lateral trunk, hip, and KFAs and displacement in shorter GCTs appear to be strategies demonstrated by faster athletes. As such, practitioners seeking to develop their athletes' cutting performance may consider coaching and instructing these aforementioned techniques (i.e., stiffer hip and knee and trunk lean towards direction of travel) for faster cutting performance.

It is worth highlighting that the strongest relationships observed between a biomechanical variable and COD performance explained approximately 50% of performance, while Stepwise multiple regression analysis revealed that exit velocity, peak RPF, PFC mean HBF, and IFPA together could explain 64% (r = 0.801, adjusted 62%, p = 0.048) of the variation in long cut completion time. While the focus of this study was to investigate solely the biomechanical determinants of COD performance, other factors such as muscular strength and power and linear speed qualities may account and explain the remaining variation in cutting performance (148, 275, 340, 410, 530). Furthermore, it must be emphasised that an athletes' ability to adopt these postures associated with faster cutting performance will also be partially influenced by their physical capacity (148, 275, 340, 410, 530). Thus, practitioners should consider developing an athlete's strength capacity (eccentric, dynamic, concentric, isometric, reactive strength) parallel to technique modification training.

5.5.2 Biomechanical determinants of injury risk

Focusing purely on cutting biomechanics associated with increased knee joint loads and thus ACL injury risk, of concern, large relationships were observed between peak KAMs and KIRMs (Table 5.1-5.2), and "higher-risk" KAM athletes also displayed greater peak KFMs which is associated with proximal anterior tibial shear (506, 631). This finding is problematic because ACL strain is amplified when a combination of high frontal and transverse knee moments are generated in combination with anterior tibial shear, compared to uniplanar loading (26, 288, 342, 424, 513). To the best of the author's knowledge, this is the first study to examine the relationship between frontal and transverse knee moments during cutting. The majority of investigations that have investigated the biomechanical determinants of injury risk during COD have primarily focused on KAMs (127, 139, 189, 228, 267, 272-274, 301, 358, 516, 519, 625), with only a limited number of studies investigating KIRMs (127, 189, 267, 593). The results from this study show a large relationship between the two aforementioned knee joint loads, and therefore it could be argued that determinants associated with greater KAMs are also likely to amplify KIRMs. Furthermore, the athletes classified as "high KAMs", although specific to the laboratory and methodological procedures adopted in the thesis, displayed peak KAMs of ~1.5 Nm/kg which were in line with previous research which have reported "excessive" and "higher risk KAM" thresholds, such as Sigward et al. (519) (1.2 ± 0.4 Nm/kg), Jones et al. (272) (~1.5 Nm/kg), and Chinnasee et al. (86) (~1.7 Nm/kg).

Stepwise multiple regression analysis revealed that FFC mean VBF and peak KAA together could explain 43% (r = 0.652, adjusted 41%, p < 0.001) of the variation in peak KAM during long 90° cutting. This finding is unsurprising because KAM is influenced by the magnitude of the GRF and moment arm, particularly in the frontal plane (301). Greater peak KAAs were moderately associated with greater peak KAMs, while low versus high KAM comparisons indicated that IC and peak KAAs were significantly greater for the higher KAM athletes with moderate to large effect sizes (Tables 5.1-5.4). This finding corroborates previous research that has demonstrated strong relationships between

KAAs and peak KAMs during COD (272, 274, 301, 358, 516). Increased KAAs at IC and abduction motion has the effect of placing the knee more medial to the resultant GRF vector and thus, increases the lever arm of the resultant GRF vector relative to the knee joint, leading to an increased KAM (272). Additionally, increases in knee valgus angle of 2° can lead to a 40 Nm change in valgus moment (359), while prospective research reported greater valgus angles were associated with increased risk of non-contact ACL injury (242). As such, reducing KAA and motion during cutting appears to be viable strategy to reduce knee joint loading, thus ACL injury risk.

A large proportion of studies have standardised and controlled approach velocity when examining the biomechanical determinants of cutting (127, 139, 267, 272-274, 358, 516, 519); however, athletes do not run at controlled approach velocities in sport. In contrast to these aforementioned studies, athletes in the present study were instructed to perform the cutting task as "fast as possible" because this will most likely reflect COD scenarios in sport. Importantly, greater frontal and transverse knee joint loads were associated with greater velocities over key instances of the PFC and FFC, thus velocity maintenance (Tables 5.1-5.4). These findings support the concept that approach velocity is a key factor regulating cutting knee joint loads (115, 148, 290, 301, 403, 574), most likely due to accumulative greater shear and impact GRFs experienced over shorter GCTs which contributes to increased knee joint loads (115, 148, 290, 403, 574). Additionally, a novel aspect of the current study were that velocity over key instances of the PFC and FFC were examined, with only one other study performing such analysis during COD using 3D motion analysis (275). The present study provides new insights into the effect of velocity on cutting knee joint loads and interestingly, velocity at FFC touch-down was the strongest determinant of knee joint loads, with a large relationship observed. Interestingly, low versus high KAM comparisons revealed low KAM athletes displayed greater reductions in velocity over the PFC (Tables 5.3-5.4). Consequently, reducing approach velocity, emphasising greater reductions in velocity over the PFC, and reducing velocity over key instances of the PFC and FFC are potential strategies to reduce knee joint loading, most likely due to the reduced shear and GRFs.

Greater FFC peak and mean braking forces were also associated with higher knee joint loads and demonstrated by high KAM athletes (Tables 5.1-5.4). These findings support previous research that also found greater braking forces were associated with greater peak KAMs (272, 273, 516, 519). Additionally, researchers have also shown greater HBFs and KFMs are associated with proximal anterior tibial shear (506, 631), both of which were demonstrated by high KAM athletes (Tables 5.1-5.4). The greater FFC braking forces and FFC KFMs demonstrated by high KAM athletes will most likely be attributed to the greater approach velocities, smaller reductions in velocity over the PFC, and potentially not displaying a PFC dominant braking strategy, as indicated by the small ES differences in peak HBF ratios (Tables 5.3-5.4). Nevertheless, as joint moments are influenced by the magnitude of the GRF and moment arm (301, 517), greater GRFs will therefore contribute to greater knee joint loads. As such, reducing GRFs in the FFC and potentially emphasising PFC dominant braking strategies (i.e., greater braking magnitudes in PFC) could be an effective strategy to reduce knee joint loads during cutting.

Technical and joint kinematic variables were also associated with greater cutting knee joint loads (Tables 5.1-5.4). For example, smaller HFA ROM was moderately to largely associated with greater knee joint loads, while high KAM athletes also displayed smaller FFC HFAs and ROM. The role of the hip during WA of high velocity tasks such as cuts and landings is a contentious issue (225, 294, 358, 458, 464, 517). Opposing the current study's findings, McLean et al. (358) found greater HFAs at IC were associated with greater peak KAMs during 45˚ cutting, while Kipp et al. (294) found rapid hip flexion during a single-leg land-and-cut was associated with greater KAMs but lower KIRMs. The authors of the aforementioned research have speculated that increasing hip flexion may compromise the ability of the medial muscle groups to support valgus loads, and that forward rotation of the trunk may affect moment arm distances and GRF vectors (294, 358).

Conversely, Hashemi et al. (225), in a narrative review, proposed that rapid co-flexion of hip and knee is a safer movement pattern, allowing the tibia and femur to harmoniously roll and glide on each other, thus reducing tibial anterior translation. Several studies have shown female athletes demonstrate smaller HFAs (more erect postures) and greater KAMs compared to males (174, 259, 359, 362, 458) during cutting, and it has been shown that increasing hip flexion and promoting a hip dominant strategy are involved GRF attenuation (511, 517, 631), energy dissipation (459, 624), reducing loading rates (517), and reducing knee joint loads (406, 459, 464, 517) during high impact tasks. Increasing hip flexion increases the moment arm distance at the hip which creates a greater external HFM (utilising hip extensor musculature), and may direct the GRF more vertically, thus reducing the KFM arm distance. This can have the effect of unloading the knee by more evenly distributing loading proximally up the lower-limb chain (274, 459, 464, 517), thus reducing the demands for the knee. Moreover, an erect trunk (i.e., less hip flexion) with the COM behind the base of support has been identified as a visual characteristic displayed during non-contact ACL injuries (509). Nevertheless, the results of this study suggest that increasing HFA and motion could be an effective strategy to reduce knee joint loads during cutting; however, due to the novelty of this finding, further research is required to confirm this in other populations.

Athletes with high KAMs displayed wider lateral foot plant distances and smaller knee flexion ROMs (Tables 5.3-5.4), both of which have been identified as key characteristics of non-contact ACL injuries (52, 63, 94, 204, 270, 291, 298, 310, 376, 426, 583) and increase the propensity to generate large frontal (127, 228, 272, 301, 593) and sagittal plane joint loads (42, 43, 343, 593, 615). A wide lateral foot plant is a major determinant of peak KAM (127, 228, 272, 301) because the GRF vector acts laterally relative to the knee, creating a greater moment arm (relative to knee joint centre) and thus, KAM. Additionally, smaller knee flexion ROMs, thus "stiffer, erect, extended" strategies, can increase KAMs (76, 593), increase impact GRFs (115, 130, 321, 640), and greater GRFs have been associated with increased KAMs (516, 519). Furthermore, extended knee positions are associated with high anterior tibial loading and shear, which can also increase ACL strain (42, 43, 343, 615). Extended knee positions are also commonly observed characteristics of non-contact ACL injury during directional changes and landing (52, 63, 94, 204, 270, 291, 298, 310, 376, 426, 583). Previous studies have shown that narrowing foot plant distances can decrease peak KAMs (126, 127) during cutting, while increasing knee flexion during cutting can reduce impact GRFs and reduce KFMs and KAMs (76, 115). Therefore, from an injury mitigation perspective, narrowing lateral foot plant distances and increasing knee flexion are recommended technical changes to promote safer COD mechanics.

5.5.3 The "performance-injury conflict" during cutting: which technique and mechanical variables should be addressed?

Despite the importance of directional changes for sports performance and its association with ACL injury risk, it is somewhat surprising that the majority of studies into COD biomechanics investigate performance (143, 275, 338, 346, 498, 530, 531, 596, 597) and injury risk determinants (117, 127, 139, 267, 272-274, 301, 358, 516, 519, 625) independently. Currently, a limited number of studies have comprehensively considered both COD performance and injury risk biomechanics (228); however, based on these studies and biomechanical principles, mechanics and techniques required for safer performance, thus injury mitigation, are at odds with performance (115, 148, 183, 228). The "performance-injury conflict" is problematic because athletes are unlikely to adopt safer strategies at the expense of faster performance (228).

The present study is the first to comprehensively investigate the biomechanical determinants of injury risk and performance in a large sample size, while also examining the PFC braking characteristics, KIRMs, and velocity profiles over the two foot contacts. Interestingly, the results of the current study support the limited research (115, 183, 228) and the concept of a "performance-injury conflict" (115, 148, 183, 228), whereby techniques and mechanics associated with faster performance (i.e., faster PFC and FFC velocity, greater FFC braking forces over short GCTs, greater KFMs, smaller

FFC hip and KFAs and ROM, wider lateral foot plants, and greater IFPAs) are in direct conflict with safer COD mechanics (i.e., reduced knee joint loading) (Tables 5.1-5.4). This issue is problematic for practitioners and athletes who want to adopt cutting strategies that maximise performance while concurrently minimising injury risk. For example, greater approach velocities and velocity over key instances of the PFC and FFC were associated with cutting faster performance and greater knee joint loads (Table 5.1 & 5.2), while athletes who demonstrated faster completion times and exit velocities also displayed greater knee joint loads (Tables 5.1-5.4). Undoubtedly, instructing athletes to perform COD actions slowly is not a viable and feasible strategy (148, 183), given its importance for faster performance (213, 275). As such, practitioners must acknowledge that increased knee joint loads are typically associated with greater approach and COD velocity profiles, and should therefore progress COD velocity progressively and cautiously with their athletes (148).

In agreement with Havens and Sigward (228), faster athletes and athletes with greater peak KAMs displayed greater peak KFMs (Tables 5.1-5.4). These increased sagittal plane knee moments are most likely a product of the faster approach velocities and braking forces. Importantly, greater frontal and transverse knee joint loads were associated with greater approach velocities, greater velocity profiles over key instances of the PFC and FFC, and velocity maintenance (Tables 5.1-5.4). These findings support the concept that approach velocity is a key factor regulating cutting knee joint loads (115, 148, 290, 301, 403, 574), most likely due to accumulative greater shear and impact GRFs experienced over shorter GCTs which contributes to increased knee joint loads (115, 148, 290, 403, 574), and the increased eccentric demands during the braking (WA) phase (37, 212, 341). In addition, higher impact and braking forces over shorter GCTs were demonstrated by faster athletes and associated with faster performance (Tables 5.1-5.4). Conversely, lower braking forces over longer GCTs were characteristics associated with lower knee joint loads but negatively impacting performance (Tables 5.1-5.4). Greater KFMs and posterior GRFs are also associated proximal anterior tibial shear (506, 631) and potential ACL loading (42, 43, 343, 615), but are also associated with faster cutting performance (146, 228) (Tables 5.1-5.4); highlighting the conflict between performance and injury. Again, braking forces and GCT are influenced by an athlete's approach velocity and therefore, given its importance for performance, lowering braking forces and increasing GCT duration are not advisable strategies for coaches to implement with their athletes but should acknowledge this conflict when coaching cutting.

From a technical perspective, sagittal plane lower-limb kinematics also have an important role for performance and injury risk (183). For example, athletes which demonstrated faster performance yet greater knee joint loads, demonstrated smaller FFC hip and knee flexion ROM and greater ankle dorsi-flexion ROM and thus, arguably a "stiffer" hip and knee strategy. In contrast, athletes who demonstrated smaller knee joint loads but slower performance, displayed greater FFC hip and knee flexion ROM and smaller ankle dorsi-flexion ROM, and therefore a potentially more compliant and softer strategy. This result supports Dai et al. (115) that found increasing knee flexion during cutting concurrently reduced braking GRFs and peak KFMs, but negatively impacted performance by increasing GCT and reducing exit velocity. Celebrini et al. (76) found increasing knee flexion reduced KAMs during cutting, while Welch et al. (597) has found resisting hip flexion over WA was associated with faster cutting performance. In some cases, faster athletes in the present study displayed 0° of hip flexion ROM (Figure 5.1a), but also tended to display greater KAMs (Figure 5.1b), though it is worth noting that the relationships between HFA ROM and exit velocity and peak KAMs appeared better (i.e. more congregated around line of best fit) with higher magnitudes of exit velocity and peak KAMs.

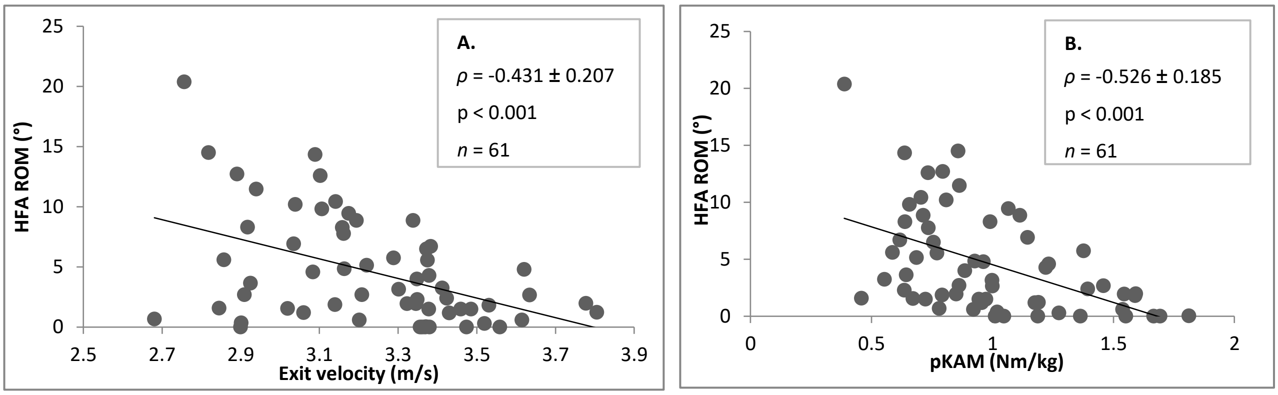


Figure 5.1. Scatter plots for HFA ROM association with short cut exit velocity and peak KAMs

The stiffer (i.e., reduced ROM) hip and knee strategy is effective for performance by reducing GCT and potentially permitting more effective reactive strength and utilisation of the SSC (55, 317, 597), thus facilitating more effective force transmission due to the rapid transition from braking (WA) to push-off. However, Hashemi et al. (225) proposed the "hip extension, knee flexion paradox" as a potential ACL injury mechanism. They suggest that limited and/or slow hip flexion relative to knee flexion (i.e., lack of co-flexion) during WA is a "high-risk" movement pattern resulting in greater tibial anterior translation, whereas active co-flexion is protective because it permits the tibia and femur to harmoniously roll and glide on each other (Figure 5.2). Furthermore, stiffer and extended braking strategies ineffectively dissipates forces and energy (115, 130, 321, 511, 624, 640), increases loading rates (517), and may increase anterior tibial (42, 43, 343, 615) and knee abductor loading (75, 406, 459, 517). Soft WA strategies are often coached in injury mitigation programmes (76, 239, 247, 432) to reduce impact GRFs and knee joint loads; however, practitioners must consider the conflict between performance and injury risk when manipulating such sagittal plane joint kinematics. Because ACL injuries occur ≤ 50 ms at extended knee postures with minimal hip flexion (297, 298), encouraging greater initially flexed postures with rapid hip and knee co-flexion could be a safer cutting strategy (225).

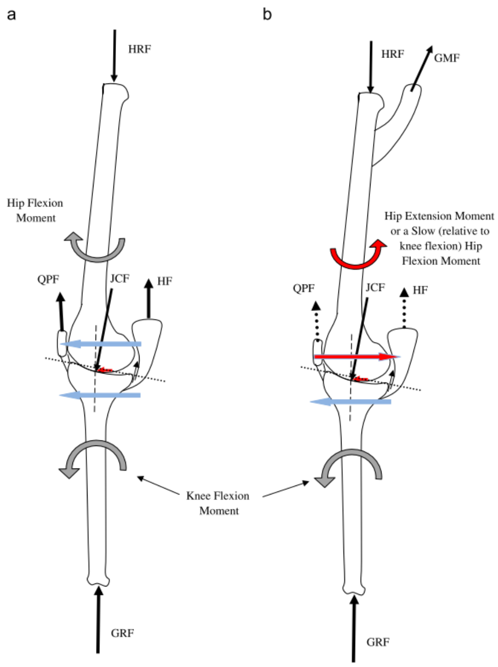


Figure 5.2. Hip extension, knee flexion paradox. (a) Normal condition at full extension, neuromuscular control assures that both knee and hip joints flex together and (b) abnormal condition under which knee is forced to flex but hip is forced to extend. HRF: hip reaction force; GMF: gluteus maximus force. (Not a free body diagram). Taken from Hashemi et al. (225)

A contentious technique variable from a performance and injury risk perspective is lateral foot plant distance (183, 228, 272). A wide lateral foot plant is associated with greater KAMs (127, 228, 272, 301), but required for faster performance and superior exit velocity due to the ML propulsive impulse requirements (11, 228, 261, 272, 597). The results of this study further substantiate the conflict between performance and injury risk with respect to lateral foot plant distance because higher knee joint loads were observed with wider lateral foot plant distances, while faster athletes tended to display wider lateral foot plant distances for superior performance and greater MLPFs (moderate to large relationships) (Tables 5.1-5.4). This result supports Havens and Sigward (228) that also found wider lateral foot plant distances were associated with faster performance and KAMs during cutting. Dempsey et al. (126, 127) has demonstrated that reducing lateral foot plant distances during cutting reduces knee joint loads; however, the researchers did not consider the implications of this technique modification on performance. Based on the results of this study and previous research (228, 597), instructing and adopting a narrower foot plant is likely to be detrimental for performance. As such, practitioners must consider the trade-off between performance and knee joint loading when coaching a wide lateral foot plant during side-step cutting.

Transverse variables pelvic rotation and IFPAs were also variables linked to faster performance and greater knee joint loads (Tables 5.1-5.4). In the present study, faster athletes demonstrated greater IFPAs and pelvic rotation towards the direction of travel; however, knee joint loads were also greater for athletes who adopted this strategy. The results of previous studies have shown greater IFPAs are associated with greater knee joint loading during COD (127, 274, 519), while rotation of the pelvis towards the intended direction of travel would result in deceleration in a transverse position, which in all probability is not as safe as a sagittal plane deceleration approach which most likely utilises the strong hip extensor musculature (76, 228, 274). Conversely, from a performance perspective, pelvis and foot rotation towards the intended direction has been suggested to facilitate effective realignment of the whole-body COM towards the desired COD (183, 229, 274). Again, practitioners must consider the conflict between performance and injury risk when coaching and screening pelvic and foot rotation during cutting.

Although several variables have been identified as factors linked to faster performance and greater knee joint loads, some variables have been shown to be associated with faster performance and lower knee joint loading or offer no associated performance benefits or detriments (Tables 5.1-5.4). For example, no associated performance benefits were found in terms of KAA and cutting performance (Tables 5.1-5.4). This result is in line with Marshall et al. (346) that demonstrated no meaningful relationship between knee valgus and completion time. Consequently, practitioners should look to minimise knee valgus in their athletes to reduce knee joint loads, with no associated performance detriments during cutting. Greater PFC HBFs and smaller braking ratios (i.e., more PFC dominant braking) were variables associated with faster performance and lower knee joint loads (Tables 5.1-5.4). This result corroborates previous research that has found greater PFC braking forces and PFC dominant braking strategies were associated with faster 180˚ COD performance (143, 147, 202, 275) and lower knee joint loads (143, 147, 202, 275). Additionally, during COD, athletes will need to reduce their momentum in order to perform the COD (213, 227, 275). As such, encouraging greater reductions in velocity over the PFC to lower the subsequent velocity at FFC (key determinant of greater knee joint loads), by facilitating effective PFC braking, should also lower knee joint loads while maintaining effective performance. This outcome should be achieved by instructing athletes to adopt a PFC dominant braking strategy in line with recently published guidelines (147). Finally, smaller lateral trunk flexion angles and ROM were associated with smaller GCTs, a critical determinant of faster performance (143, 338, 346, 498, 531, 596, 597), while greater lateral trunk flexion angles have been shown to increase knee joint loads (127, 267, 272). Previous research has shown that medial trunk lean towards the direction of travel was associated with faster performance (346, 597). Consequently, practitioners should look to instruct cutting techniques with smaller lateral trunk flexion (trunk lean towards the intended cut) and ROM for faster and safer performance.

Overall, mechanics associated with faster cutting performance are in direct conflict with mechanics for safer cutting. It is important to note that optimal performance and "high-risk" knee joint loading is not attributed solely to one variable, but the amalgamation and interaction of velocity, joint kinematics and kinetics, and braking and propulsive force. As such, practitioners must consider the performance and injury risk implications when coaching and modifying cutting techniques. In light of the finding that faster athletes generally display greater knee joint loads and are unlikely to sprint slower, it is imperative that athletes have the physical capacity (i.e., neuromuscular control, co-contraction, and rapid force production) and technique to tolerate the knee joint loading demands of side-steps (39, 272, 301, 328, 410, 432, 544, 591). Moreover, given the importance of velocity for faster performance, it is integral that practitioners progressively expose athletes to cutting drills of higher velocity (148) and consider the athlete's training status and strength capacity when exposing them to high velocity cutting drills (131, 409, 410).

5.6 LIMITATIONS

It is worth noting that there were several limitations in the present study. Firstly, males were only investigated, thus caution is advised regarding the generalisation of these results to female athletes. The majority of COD biomechanical determinant studies have investigated female athletes, and there is a lack of data investigating male athletes, thus rationalising the importance of investigating this sex (6, 7). Secondly, the biomechanical demands are angle-dependent (39, 108, 212, 213, 227, 229, 504, 505, 516), thus the findings of this study are applicable to 90° cuts only. Further insight is required into biomechanical determinants of performance and injury risk in cuts and turns of different angles and actions. Thirdly, although a standardised surface was used, this surface does not reflect the grass and artificial field-turfs that the athletes regularly perform their CODs on. Additionally, discrete data analysis was carried out, thus peak knee joint moments were only examined. It was unclear, however, if the peak KAMs and KIRMs coincided with the peak sagittal plane lower-limb moments, postures, and GRF. Thus, future research is necessary that considers the full temporal waveform for further insight into the biomechanical determinants of cutting performance and knee joint moments. It should also be noted that the athletes intended to perform a 90° cut (70-90° range) but the actual true cutting angle performed by the athletes was ~60°. Therefore, this suggests that the step after the lateral foot plant is involved in redirecting the athlete.

5.7 CONCLUSION

In conclusion, the results of this study confirm that techniques and mechanics associated with faster cutting performance (i.e., faster velocities over key instances of the PFC and FFC, greater FFC braking forces over short GCTs, greater FFC KFMs, smaller hip and KFAs and ROM, wider lateral foot plants, and greater IFPAs) are in direct conflict with safer cutting mechanics (i.e., reduced knee joint loading) (Tables 5.1-5.4), and support the "performance-injury conflict" concept during cutting (115, 148, 183). Consequently, practitioners must be cautious when coaching and manipulating cutting technique and mechanics, and acknowledge the implications of technique modification on performance and potential injury risk. Because athletes are driven by performance, techniques and mechanics that result in effective performance even at the expense of greater knee joint loading will inevitably be adopted and will also be a by-product of their sport. As such, practitioners should look to develop their athletes' physical capacity (i.e., neuromuscular control, co-contraction, and rapid force production) and technique to tolerate and support the knee joint loading demands of side-steps (39, 272, 301, 328, 341, 410, 432, 544, 591). Knee valgus is linked with greater knee joint loads with no associated performance benefits, while PFC braking dominant strategies and minimising lateral trunk flexion are factors associated with faster performance and safer COD mechanics. Therefore, coaching PFC dominant braking strategies and minimising knee valgus and lateral trunk flexion should facilitate effective performance and reduce knee joint loading.

CHAPTER 6: The Cutting Movement Assessment Score (CMAS): A Qualitative Screening Tool to Identify Athletes With "High-Risk" Movement Mechanics During Cutting

6.1 INTRODUCTION

Side-step lateral foot plant-and-cut actions are frequently performed movements in numerous sports (171, 184, 409, 499, 510, 601) and are also linked to decisive moments, such as evading an opponent to penetrate the defensive line in rugby (tackle-break success in rugby) (601), or getting into space to receive a pass in netball (184). Side-step cutting, however, is an action associated with non-contact ACL injuries in sports (52, 63, 94, 170, 270, 291, 298, 376, 426, 583). Although ACL injury risk factors are multifactorial (469) and a complex interaction of internal and external factors (i.e., anatomical, hormonal, biomechanical, neuromuscular, environmental) (48, 241, 310), a large proportion of ACL injuries are non-contact in nature during high velocity and impact sporting tasks, such as side-stepping (52, 270, 310). This occurrence can be attributed to the tendency to generate large multiplanar knee joint loading, such as KAMs and KIRMs (39, 126, 127, 273, 301), which increase ACL strain (26, 288, 342, 424, 513). These potentially hazardous knee joint loads are amplified when poor initial postures and movement is demonstrated (biomechanical and neuromuscular control deficits) during cutting (183, 241, 245, 246, 391, 508), but importantly these deficits are modifiable (241, 432). As such, understanding the mechanics, interventions, and screening tools that can assist in ACL injury risk mitigation is of critical importance.

The ability to identify athletes potentially "at-risk" of injury is a critical step in effective ACL injury risk mitigation (185, 241). Although it is inconclusive whether screening tools can predict non-contact ACL injury (18, 187, 353), evaluating movement quality and identifying biomechanical and neuromuscular control deficits (high-risk movement patterns) can provide important information regarding an athlete's "injury risk profile" (240, 354, 372). These "high-risk" deficits include KAA (272, 274, 301, 358, 516), lateral trunk flexion (127, 141, 189, 267, 272), extended knee postures (115, 298, 593), and hip internal rotation (228, 358, 516, 519). This information from movement screening can subsequently be used to inform the future prescription of training and conditioning so that specific deficits can be targeted through appropriate training interventions to decrease the relative risk of injury (240, 243, 372). Therefore, the inclusion of valid and reliable screening tools that assess movement quality are an important component of sports medicine and strength and conditioning testing batteries to provide an "injury risk profile" for an athlete (240, 278).

3D motion analysis is considered the gold standard for evaluating movement kinetics and kinematics (187, 241); however, this method can be susceptible to errors, with a diverse range of data collection and analysis procedures available to practitioners (e.g., markers sets, marker placement, soft tissue artefact, body segment inertial parameters, modelling, reference frame, normalisation, low-pass filtering COFs) which can affect the outcome values, reliability, or subsequent evaluations of an athlete's biomechanical profile (70, 303-305, 371, 417, 488, 489). Given these methodological considerations and issues, and the fact the 3D motion analysis is expensive, time-consuming, requires expert and well trained assessors, and is usually restricted to testing one participant in a laboratory setting, time- and cost-effective qualitative field-based screening tools have been developed, such as the LESS (430, 434), TJA (238, 397), and QASLS (8, 237), to assess lower-limb and whole-body postures associated with increased potential risk of injury (high-risk movement patterns). However, to the best of the author's knowledge, the LESS is the only screening tool that has been validated against 3D motion analysis (427, 434).

It is worth noting that the LESS, TJA, and QASLS generally assess landing mechanics during a vertical-orientated task. Although screening landing mechanics is indeed applicable to jump-landing sports (netball, basketball, volleyball) where the primary action associated with non-contact ACL injury is landing manoeuvres (54, 248, 310, 541), these aforementioned assessments may lack specificity to the unilateral, multiplanar plant-and-cut manoeuvres observed when changing direction (187, 278, 372). This is particularly important when aiming to screen athletes who participate in sports such as soccer (583), handball (426), American football (270), badminton (291), and rugby (376), where directional changes are a primary action associated with non-contact ACL injury. Furthermore, there are mixed findings as to whether examination of landing mechanics can identify athletes with poor cutting mechanics (4, 86, 302, 421), with evidence suggesting an athlete's mechanics and "injury risk profile" are task dependent (86, 279, 302, 388). As such, screening side-step cutting technique, which is specific to the actions associated with non-contact ACL injuries in cutting sports (i.e., rugby, handball, soccer, American football), could be a more effective strategy for identifying poor cutting movement quality in athletes, which can help inform future injury risk mitigation training.

Unfortunately, there is a paucity of field-based cutting screening tools available for practitioners. McLean et al. (361) initially evaluated 2D estimates of frontal plane knee motion during cutting against the gold standard of 3D, and found 2D estimates correlated well with side-step (r^2 = 0.58) and side-jump (r^2 = 0.64) 3D valgus angles, but poorer associations were observed with 180° turn knee valgus angle (r^2 = 0.04); thus, highlighting the difficulty in assessing 2D valgus motion in the frontal plane using a single camera during sharp CODs. Hamdan et al. (218) reported reliable assessments of 2D sagittal plane hip and knee angles during 45°side-step cutting, but did not validate against 3D motion analysis or assess frontal plane kinematics. Recently, Weir et al. (593) has demonstrated that 2D measures of dynamic knee valgus angle, KFA at foot-strike and ROM, and trunk flexion ROM, when inserted in regression equations, can be used to predict 3D peak KFMs, KAMs and KIRMs during unanticipated 45° side-steps. Despite these promising relationships, 2D side-step screening methods are not widely adopted by practitioners and clinicians. The lack of adoption could be attributed to the 2D method requiring additional time and software to measure joint kinematics (digitisation), thus potentially limiting its applicability in field-settings.

In light of the issues associated with 2D analysis, Jones et al. (278) have recently developed the CMAS, which is a qualitative screening tool that assesses cutting movement quality and specific lower-limb and trunk characteristics that are associated with peak KAMs (127, 189, 228, 267, 272, 273, 301, 358, 516, 519, 593) (Appendix 5.1), such as PFC braking strategy, and trunk, hip, knee, and foot positioning and motions. In a preliminary study (278), a strong relationship between CMAS and peak KAM (ρ = 0.633; $p < 0.001$) was demonstrated, while moderate to excellent intra- and inter-rater agreements for all CMAS variables (intra-rater: k = 0.60-1.00, 75-100% agreements; inter-rater: k = 0.71-1.00, 87.5-100% agreements) were observed, although lower inter-rater agreements for trunk positioning were observed (k = 0.40, 62.5% agreement). In light of these findings, the CMAS may have the potential to identify athletes displaying "high-risk" cutting mechanics but importantly, could be used as a technical framework for coaching safer cutting mechanics. It should be noted, however, that the preliminary study contained a small sample size (n = 8 participants, 36 trials) and must be expanded with a greater sample size to confirm its validity and reliability. Furthermore, the authors recommended an additional camera to be placed at 45° relative to the COD and using a higher video capture rate (≥100 Hz) to permit more accurate and reliable assessments for frontal and transverse plane technique deficits (i.e., trunk positioning, knee valgus).

The aim of this study, therefore, was to assess the validity of the CMAS tool to estimate the potential peak KAMs against the gold standard of 3D motion analysis, expanding on the work of Jones et al. (278) by examining a larger sample size and using an additional camera recording at a higher sampling rate. A further aim was to determine whether "higher-risk" movement mechanics were displayed by participants with higher CMASs compared to participants with lower CMASs. Firstly, it was hypothesised that excellent inter- and intra-rater reliability would be demonstrated for CMAS items. Secondly, in line with Jones et al. (278), it was hypothesised that a strong relationship would be demonstrated between CMAS and peak KAM, and the CMAS would be able to discriminate between "low" and "high" CMASs in terms of "high-risk" whole-body kinetics and kinematics.

6.2 METHODS

6.2.1 Experimental approach

This study used a mixed, cross-sectional design to determine the relationship between CMAS and peak KAMs during cutting over one session, following an associate strategy. A between-subject, comparative design was used to compare cutting mechanics between CMAS trials and participants with "low" and "high" CMASs. Participants performed six 90° cuts whereby 3D motion and 2D video footage data were simultaneously captured to permit qualitative screening and comparisons to 3D motion data, similar to the procedures of previous research (278, 434).

6.2.2 Participants

Based on the work of Jones et al. (278) who determined the relationship between CMAS and peak KAM, a minimum sample size of 29 was determined from an *a priori* power analysis using G*Power (Version 3.1, University of Dusseldorf, Germany) (172). This was based upon a correlation value of ρ = 0.633, a power of 0.95, and type 1 error or alpha level of 0.05. As such, 41 athletes (28 males/13 females) from multiple sports (soccer n = 20, rugby n = 5, netball n = 9, and cricket n = 7) (mean ± SD; age: 21.3 ± 4.0 years, height: 1.75 ± 0.08 m, mass: 72.8 ± 11.8 kg) participated in this study. For inclusion in the study, all athletes had played their respective sport for a minimum of five years and regularly participated in one game and performed two structured skill-based training sessions per week. All athletes were free from injury and had never suffered a prior traumatic knee injury such as an ACL injury. At the time of testing, players were currently in-season (competition phase). The investigation was approved by the Institutional Ethics Review Board (HSR1617-02), and all participants were informed of the benefits and risks of the investigation prior to signing an institutionally approved consent or parental assent document to participate in the study.

6.2.3 Cutting Movement Assessment Score

Table 6.1 presents the CMAS qualitative screening analysis tool to estimate the magnitude of KAMs during cutting, which has been slightly modified from the preliminary investigation by Jones et al. (278) (i.e., extra description provided to some criteria). The CMAS is based on research pertaining to technical determinants of peak KAMs during 30-90° side-step cutting (Figure 5.1) (127, 189, 228, 267, 272, 273, 301, 358, 516, 519, 593), visual observations of non-contact ACL injuries (270, 298, 426), and findings from Chapter 5 (Study 2), whereby higher KAM athletes tended to display wider lateral foot plant distances, greater KAAs, smaller knee and hip flexion ROMs, greater vertical GRFs, and lower PFC braking forces. Appendix 5.1 contains operational definitions and a biomechanical rationale of the

CMAS. If an athlete exhibits any of the characteristics in Table 6.1 they are awarded a score, with a higher score representative of poorer technique and potentially greater peak KAM (278).

Table 6.1. Cutting Movement Assessment Score tool

Camera	**Variable**	**Observation**	**Score**
Penultimate contact			
Side / 45°	**Clear PFC braking strategy** (at initial contact) • Backward inclination of the trunk • Large COM to COP position – anterior placement of the foot • Effective deceleration – heel contact PFC	**Y/N**	**Y=0/ N=1**
Final Contact			
Front / 45°	**Wide lateral leg plant** (approx. > 0.35 m – dependent on participant anthropometrics) (at initial contact)	**Y/N**	**Y=2/N=0**
Front / 45°	**Hip in an initial internally rotated position** (at initial contact)	**Y/N**	**Y=1/N=0**
Front / 45°	**Initial knee 'valgus' position** (at initial contact)	**Y/N**	**Y=1/N=0**
All 3	**Foot not in neutral foot position** (at initial contact) Inwardly rotated foot position or externally rotated foot position (relative to original direction of travel)	**Y/N**	**Y=1/N=0**
Front / 45°	**Frontal plane trunk position relative to intended direction; Lateral or trunk rotated towards stance limb, Upright, or Medial** (at initial contact and over WA)	**L/TR/U/M**	**L/TR=2/ U = 1, /M=0**
Side / 45°	**Trunk upright or leaning back throughout contact** (not adequate trunk flexion displacement) (at initial contact and over WA)	**Y/N**	**Y=1/N=0**
Side / 45°	**Limited Knee flexion during final contact (stiff) ≤ 30˚** (over WA)	**Y/N**	**Y=1/N=0**
Front / 45°	**Excessive Knee 'valgus' motion during contact** (over WA)	**Y/N**	**Y=1/N=0**
		Total Score	**0 /11**

Key: PFC: Penultimate foot contact; COM: Centre of mass; COP: Centre of pressure; WA: Weight acceptance; TR: Trunk rotation; Y: Yes; N: No; L: Lateral; U: Upright; M: Medial.

6.2.4 Procedures

For a detailed description of the warm-up, short 90° cut, marker placement, and 3D motion analysis procedures, please refer to Chapter 3.2 and 3.3 in the research methodology. In brief, participants performed six trials of a 90° cut "as fast as possible" (Figure 6.1). Completion time (2.11 ± 0.14 seconds, CV% = 2.71%) was measured to standardise performance between trials and was assessed using two sets of Brower timing lights placed at hip height (Draper, UT, USA). Marker and force data were collected over the PFC and FFC of the cut.

Briefly, the following kinetics and kinematics were examined to provide insight into potentially "high-risk" cutting mechanics: vertical and horizontal GRF, knee flexion, rotation, and abduction angles and moments, hip rotation angle, trunk inclination angle, lateral foot plant distance, lateral trunk flexion, IFPA. These aforementioned kinetic and kinematics were evaluated because they have been shown to be associated with greater multiplanar knee joint loads (183, 301, 593), and have also been identified as visual characteristics of non-contact ACL injury during cutting (270, 298, 426). Approach velocities were 4.5 ± 0.5 m/s, by calculating the horizontal COM velocity using the combined lower-limb and trunk model. A more detailed rationale for investigation of these variables is presented in Appendix 5.1.

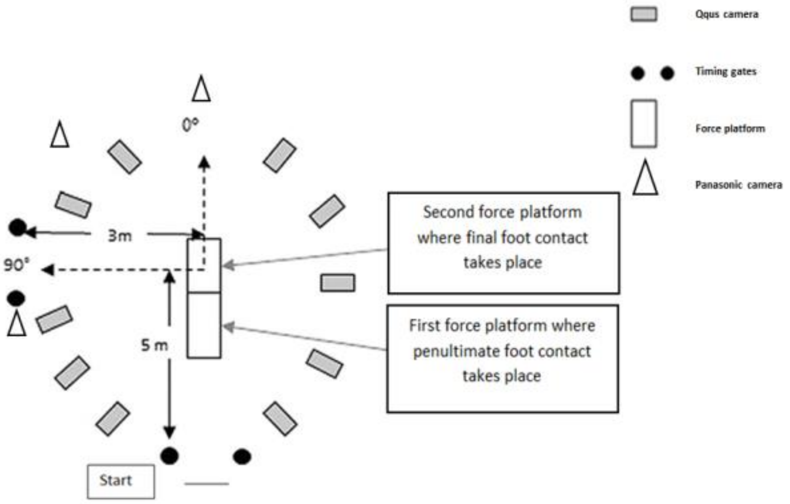


Figure 6.1. Plan view of the experimental set-up. The task involved participants approaching 5-m towards turning point on 2nd force platform. At the turning point, participants cut to the left 90° using their right limb between timing gates placed 3-m away. Marker, GRF, and 2D camera data were collected simultaneously.

<u>6.2.5 Qualitative assessment: CMAS</u>

While marker and GRF data were collected, three Panasonic Lumix FZ-200 high-speed cameras sampling at 100 Hz simultaneously filmed the cutting trials. These cameras were positioned on tripods 3-m away from the force plates at a height of 0.60-m and were placed in the sagittal and frontal plane, with a camera also placed 45° relative the cut, in accordance with previous recommendations (278) (Figure 6.1). Video footage was subsequently viewed in Kinovea software (0.8.15 for Windows, Bordeaux, France), which is free, and was used for qualitative screening using the CMAS (Table 6.1). This software allowed videos to be played at various speeds and frame-by-frame. The three raters were allowed to independently watch the videos as many times as necessary (182, 427), at whatever speeds they needed to score each test, and could also pause footage for evaluative purposes (182). On average, qualitative screening of one trial took ~3 minutes.

Prior to qualitative screening, all raters attended a one-hour training session outlining how to grade the cutting trials using the CMAS, and to establish and uniformly agree on "low-risk" and "high-risk" movement patterns using pilot video footage. Subsequently, the lead researcher created a manual for all raters which contained guidelines, operational definitions (Appendix 5.1 and 5.2), and example images of "low-risk" and "high-risk" motions of each screening criteria to assist CMAS screening.

6.3 STATISTICAL ANALYSES

Thirty-two trials were discarded due to technical issues with camera footage, 3D data, or participants who slid or missed the platform which went unnoticed during data collection, thus resulting in 214 trials (minimum 4 trials from 41 athletes) screened and used for further analysis. All statistical analyses were performed in SPSS v 24 (SPSS Inc., Chicago, IL, USA) and Microsoft Excel (version 2016, Microsoft Corp., Redmond, WA, USA). To determine inter- and intra-rater reliability, 41 trials (one trial from each participant) were randomly selected by the lead researcher, similar to the procedures of previous research (278). The lead researcher, who has seven years' strength and conditioning and biomechanics experience, viewed and graded each trial on two separate occasions separated by seven days, in line with previous research (182, 434) to examine intra-rater reliability. Another researcher (experienced biomechanist; 17 years' biomechanics and strength and conditioning experience), viewed and graded each trial once and these scores were compared to the lead researcher to establish inter-rater reliability. In addition, a recent Sports Science graduate also viewed and graded each trial once and these scores were compared to the lead researcher to establish inter-rater reliability.

ICCs (two-way mixed effects, average measures, absolute agreement) for total score were determined and were interpreted based on the following scale presented by Koo and Li (299): poor (< 0.50), moderate (0.50-0.75), good (0.75-0.90), and excellent (> 0.90). For each item within the CMAS (Table 6.1), percentage agreements (agreements /agreements + disagreements × 100) and Kappa coefficients were calculated. Kappa coefficients were calculated using the formula; k = Pr(a) – Pr(e) / 1 – Pr(e), where Pr(a) = relative observed agreement between raters; Pr(e) = hypothetic probability of chance agreement, which describes the proportion of agreement between the two methods after any agreement by chance has been removed (577). The kappa coefficient was interpreted based on the following scale of Landis and Koch (314): slight (0.01-0.20), fair (0.21-0.40), moderate (0.41-0.60), good (0.61-0.80), and excellent (0.81-1.00). Percentage agreements were interpreted in line with previous research (107, 427) and the scale was as follows: excellent (≥ 80%), moderate (51-79%), and poor (≤ 50%) (107, 427).

The relationship between CMAS and the "gold standard" determination of peak KAM during the cutting task from 3D motion and GRF analysis using the means of each participant was explored using Spearman's rank correlation, with 95% CIs, due to the non-parametric nature of the qualitative data. Correlations were evaluated as follows: trivial (0.00 - 0.09), small (0.10 – 0.29), moderate (0.30 – 0.49), large (0.50 – 0.69), very large (0.70 – 0.89), nearly perfect (0.90 – 0.99), and perfect (1.00) (255). This analysis was performed using the 214 trials screened by the lead researcher.

Participants were classified into low CMAS (bottom 33%, $n = 14$) and high CMAS (top 33%, $n = 14$) groups based on their mean CMASs. Subsequently, cutting 3D kinetics and kinematics were compared between the two groups (participant mean data) using independent sample t-tests for parametric data and Mann-Whitney U tests for non-parametric data. To explore the magnitude of differences between groups, mean differences and Hedges' *g* effect sizes with 95% CIs were also calculated as described previously (231) (Chapter 4.4), and interpreted as trivial (≤ 0.19), small (0.20–0.59), moderate (0.60–1.19), large (1.20–1.99), very large (2.00–3.99), and extremely large (≥ 4.00) (255). Statistical significance was defined as $p \leq 0.05$ for all tests.

Additionally, the relationship between CMAS and peak KAMs was assessed for all pooled trials using Spearman's correlations. The 214 trials were classified into low (CMAS ≤ 3), moderate (CMAS 4-6), and high (CMAS ≥ 7) groups based on pooled CMAS mean–1 SD considered low, mean+1 SD considered high, and mean±1 SD considered moderate. Subsequently, cutting 3D kinetics and kinematics were compared between the three groups using a one-way ANOVA for parametric data, with Bonferroni post-hoc comparisons in cases of significant differences. Kruskal-Wallis tests were used for non-parametric data, with Mann-Whitney U pairwise comparisons in cases of significant differences. To explore the magnitude of differences between groups, Hedges' *g* effect sizes with 95% CIs were calculated as described above.

6.4 RESULTS

6.4.1 Intra- and inter-rater reliability

Excellent intra-rater reliability was observed for CMAS total score (ICC = 0.946). Intra- and inter-rater percentage agreements and Kappa coefficients are presented in Table 6.2. Excellent intra-rater percentage-agreements and kappa coefficients were demonstrated for all CMAS variables (Table 6.2), with two variables scoring 100% agreement. For inter-rater reliability, most items displayed moderate to excellent percentage agreements (Table 6.2), while most items displayed moderate to good kappa coefficients between the lead researcher and experienced biomechanist. Conversely, kappa coefficients ranged from slight to good between the lead researcher and recent graduate, and most items displayed moderate to excellent percentage agreements (Table 6.2). Moderate inter-rater reliability was observed for CMAS total score between raters (ICC = 0.690).

Table 6.2. Intra- and inter-rater reliability for CMAS criteria

Variable/ CMAS tool criteria	Intra-rater reliability (Lead researcher)		Inter-rater reliability - Lead researcher vs. experienced biomechanist		Inter-rater reliability - Lead researcher vs. recent graduate	
	% agreement	***k***	**% agreement**	***k***	**% agreement**	***k***
Clear PFC braking	97.6	0.940	82.9	0.633	82.9	0.633
Wide lateral leg plant	95.1	0.900	82.9	0.629	87.8	0.747
Hip in an initial internally rotated position	100.0	1.000	63.4	0.194	43.9	0.067
Initial knee 'valgus' position	90.2	0.805	75.6	0.512	75.6	0.512
Inwardly rotated foot position	100.0	1.000	80.5	0.599	90.2	0.784
Frontal plane trunk position relative to intended direction	90.2	0.805	73.2	0.551	87.8	0.767
Trunk upright or leaning back throughout contact	100.0	1.000	90.2	0.554	78.0	0.220
Limited Knee Flexion during final contact	97.6	0.932	80.5	0.431	80.5	0.381
Excessive Knee 'valgus' motion during contact	95.1	0.898	80.5	0.605	70.7	0.376
Average	96.2	0.920	78.9	0.520	77.5	0.500

Key: CMAS: Cutting movement assessment score; PFC: Penultimate foot contact

6.4.2 Relationships between CMAS and peak KAM (participant average data)

Mean ± SD from each trial of the 41 participants were 5.1 ± 1.8 CMAS and peak KAM 1.00 ± 0.44 Nm/kg. CMASs and KAMs for males and females were 5.1 ± 1.7, 1.07 ± 0.45 Nm/kg and 5.2 ± 2.1, 0.81 ± 0.35 Nm/kg, respectively. Figure 6.2 shows a linear and positive relationship between CMAS and peak KAMs. Spearman's correlation revealed a significant and very large (ρ = 0.796, 95% CI = 0.647-0.887, p < 0.001) association between CMAS and peak KAMs (Figure 6.2).

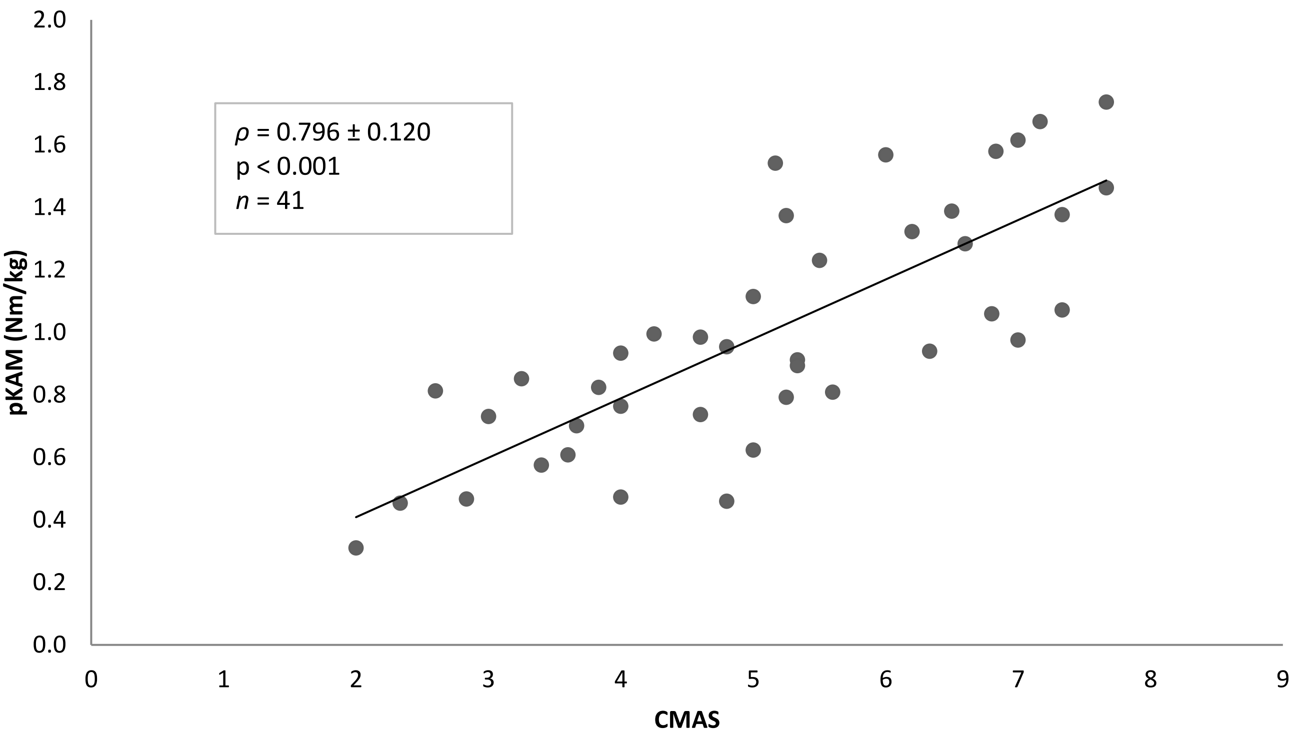


Figure 6.2. **Relationship between CMAS and peak KAMs (pKAM) participant mean data**

6.4.3 Comparisons in 3D cutting kinetics and kinematics between participants with "low" and "high" CMASs (participant average data)

Descriptive statistics, *p* values, and effect sizes for kinetic and kinematic measures for participants with "low" and "high" CMASs are presented in Table 6.3. Participants with higher CMASs displayed significantly greater FFC mean VBFs, HBFs, and mean HBF ratios, and greater peak KAAs, IFPAs, and greater lateral foot plant distances (Table 6.3), with moderate to large effect sizes. Additionally, significantly greater cutting multiplanar knee joint loads (KFMs, KIRMs, and KAMs) were demonstrated by participants with higher CMASs compared to lower (Table 6.3), with moderate to very large effect sizes.

Table 6.3. Comparison of 3D cutting mechanics between participants with lower and higher CMASs containing *p* values and effect sizes

	Variable	Foot contact	Low CMAS (*n* = 14)		High CMAS (*n* = 14)		*p*	*g*	95% CI		Mean difference	Mean difference 95% CI	
			Mean	SD	Mean	SD			LB	UB		LB	UB
	CMAS		3.34	0.70	6.95	0.63	**<0.001****	-5.29	-6.87	-3.72	-3.61	-4.13	-3.10
GRF	**peak VBF (BW)**	**PFC**	2.67	0.55	2.72	0.63	0.855	-0.07	-0.81	0.67	-0.04	-0.50	0.42
	mean VBF (BW)	**PFC**	0.95	0.16	0.97	0.20	0.879	-0.06	-0.80	0.68	-0.01	-0.15	0.13
	peak HBF (BW)	**PFC**	-1.53	0.52	-1.50	0.48	0.872	-0.06	-0.80	0.68	-0.03	-0.42	0.36
	mean HBF (BW)	**PFC**	-0.56	0.12	-0.53	0.14	0.617	-0.18	-0.92	0.56	-0.02	-0.12	0.07
	peak VBF (BW)	**FFC**	2.55	0.53	2.64	0.46	0.632	-0.18	-0.92	0.56	-0.09	-0.48	0.30
	mean VBF (BW)	**FFC**	1.54	0.18	1.71	0.21	**0.029***	-0.84	-1.61	-0.07	-0.17	-0.33	-0.02
	peak HBF (BW)	**FFC**	-1.44	0.35	-1.45	0.24	0.975	0.02	-0.73	0.76	0.00	-0.23	0.23
	mean HBF (BW)	**FFC**	-0.78	0.16	-0.94	0.13	**0.009***	1.03	0.24	1.82	0.16	0.04	0.27
	peak HBF ratio	**both**	1.03	0.35	1.06	0.39	*0.909*	-0.09	-0.83	0.66	-0.03	-0.32	0.26
	mean HBF ratio	**both**	1.42	0.29	1.88	0.65	***0.018****	-0.88	-1.66	-0.10	-0.45	-0.84	-0.06
Joint kinematics	**peak KFA (°)**	**FFC**	66.6	9.0	62.5	7.5	0.209	0.47	-0.28	1.22	4.0	-2.4	10.5
	KFA - IC (°)	**FFC**	23.1	5.1	23.6	4.9	0.766	-0.11	-0.85	0.63	-0.6	-4.5	3.3
	KFA ROM (°)	**FFC**	43.5	7.3	38.9	5.9	0.080	0.67	-0.09	1.43	4.6	-0.6	9.8
	peak KAA (°) (- abduction, + adduction)	**FFC**	-7.8	6.5	-13.4	6.6	**0.032***	0.83	0.06	1.60	5.6	0.5	10.7
	KAA - IC (°) (- abduction, + adduction)	**FFC**	4.3	4.8	0.6	4.7	0.052	0.75	-0.02	1.51	3.7	0.0	7.4
	KAA ROM (°)	**FFC**	-12.1	4.9	-14.0	5.4	0.321	0.37	-0.38	1.12	2.0	-2.0	5.9
	KRA - IC (°) (- internal, + external)	**FFC**	-10.7	6.9	-4.5	6.2	**0.020***	-0.91	-1.69	-0.13	-6.2	-11.3	-1.1
	peak KRA (°) (- internal, + external)	**FFC**	-9.6	7.4	-1.0	8.6	**0.009***	-1.04	-1.83	-0.25	-8.6	-14.8	-2.3
	HRA - IC (°) (- internal, + external)	**FFC**	11.0	7.1	7.9	10.6	0.377	0.33	-0.42	1.08	3.1	-3.9	10.1
Technique	**Trunk inclination angle - IC (°) (relative to vertical line, + forward, - backward)**	**PFC**	6.8	3.9	8.1	3.4	0.361	-0.34	-1.09	0.41	-1.3	-4.1	1.6
	Trunk inclination angle - IC (°) (relative to vertical line, + forward, - backward)	**FFC**	9.1	5.9	10.4	6.0	0.553	-0.21	-0.96	0.53	1.3	-6.0	3.3
	IFPA - IC (°) (+ internal, - external)	**FFC**	9.0	10.2	25.5	9.3	**<0.001****	-1.64	-2.49	-0.78	-16.5	-24.1	-8.9
	Lateral trunk flexion - IC (°) (- over stance leg, + direction of travel)	**FFC**	-18.4	8.0	-17.6	7.3	0.794	-0.10	-0.84	0.64	-0.8	-6.7	5.2
	Lateral foot plant distance - IC (m)	**FFC**	-0.299	0.041	-0.336	0.044	**0.028***	0.85	0.08	1.63	0.038	0.004	0.071
Joint moment	**peak KFM (Nm/kg)**	**FFC**	3.06	0.60	3.64	0.72	**0.027***	-0.86	-1.64	-0.09	-0.59	-1.10	-0.07
	peak KRM (Nm/kg) (- internal, + external)	**FFC**	-0.69	0.39	-1.10	0.61	**0.047***	0.77	0.01	1.54	0.41	0.01	0.81
	peak KAM (Nm/kg) (+ abduction, - adduction)	**FFC**	0.73	0.27	1.37	0.28	**<0.001****	2.24	-3.18	-1.29	-0.63	-0.85	-0.42

Key: VBF: Vertical braking force; HBF: Horizontal braking force; FFC: Final foot contact; PFC: Penultimate foot contact; IC: Initial contact; BW: Body weight; KFA: Knee flexion angle; ROM: Range of motion; KAA: Knee abduction angle; KRA: Knee rotation angle; IFPA: Initial foot progression angle; KFM: Knee flexor moment; KRM: Knee rotation moment; KAM: Knee abduction moment; ES: Effect size; CMAS: Cutting movement assessment scores; CI: Confidence interval; LB: Lower bound; UB: Upper bound; HRA: Hip rotation angle; *: $p \leq 0.05$; **: $p \leq 0.001$; italic denotes non-parametric.

Trivial ES (≤ 0.19) | **Small ES (0.20-0.59)** | **Moderate ES (0.60-1.19)** | **Large ES (1.20-1.99)** | **Very Large ES (2.00-3.99)** | **Extremely large ES (≥4.00)**

6.4.4 Relationships between CMAS and peak KAM (pooled data)

Spearman's correlations revealed significant, large associations between CMAS and peak KAMs when all trials were pooled (Figure 6.3a) and when examined for female trials (Figure 6.3b). A very large and significant association was observed between CMAS and peak KAMs for male trials (Figure 6.3c).

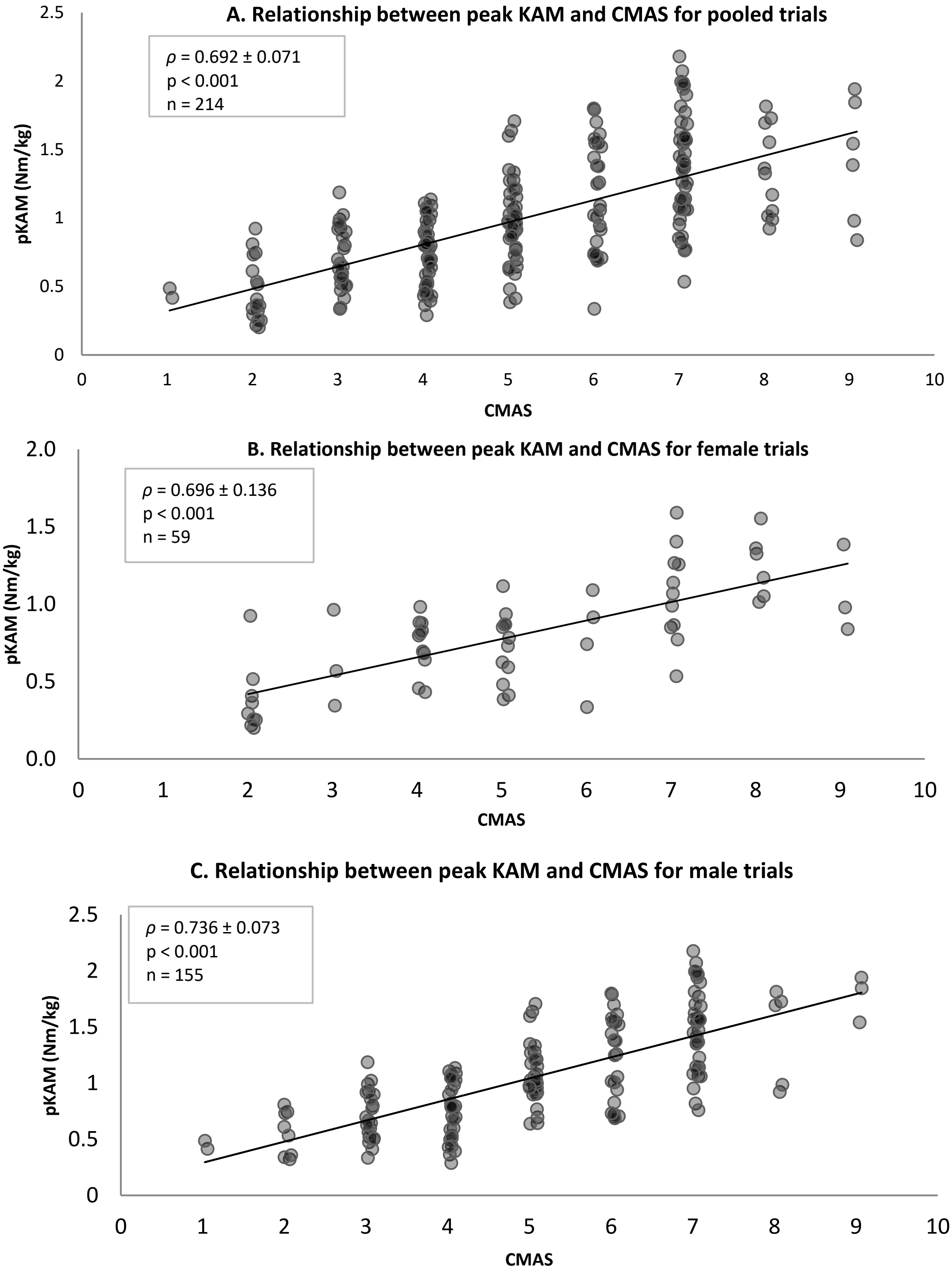


Figure 6.3. Relationship between CMAS and peak KAMs for pooled (A), female (B), and male (C) trials

6.4.5 Comparison of 3D cutting kinetics and kinematics between low, moderate, and high CMAS trials (pooled data)

Descriptive statistics, *p* values, and effect sizes for kinetic and kinematic measures for low, moderate, and high CMAS trials are presented in Tables 6.4 and 6.5. One-way ANOVAs and Kruskal-Wallis tests revealed significant differences in kinetics and kinematics between CMAS groups for most variables (Table 6.4). In general, trials performed by the participants with higher CMASs typically displayed greater peak and mean FFC VBFs, FFC HBFs, peak and mean HBF ratios, greater peak KAAs, KAA ROM, KAAs at IC, and lower peak KFA and ROM (Tables 6.4 & 6.5). Additionally, in general, greater IFPAs, greater lateral trunk flexion, greater lateral foot plant distances, and greater multiplanar knee joint moments (peak KFMs, KIRMs, and KAMs) were displayed during trials with higher CMASs compared to low, with effect sizes indicating in most cases small to large differences (Tables 6.4 & 6.5).

Table 6.4. 3D cutting mechanics descriptives for low, moderate, and high CMAS trials

	Variable	Foot contact	Low CMAS ≤ 3 (n = 43)		Moderate CMAS 4-6 (n = 109)		High CMAS ≥ 7 (n = 62)		One-way ANOVA/ Kruskal-Wallis
			Mean	SD	Mean	SD	Mean	SD	
GRF	peak VBF (BW)	PFC	2.65	0.59	_2.76_	0.66	2.80	0.71	_0.285_
GRF	mean VBF (BW)	PFC	0.95	0.19	1.00	0.18	0.97	0.22	_0.323_
GRF	peak HBF (BW)	PFC	_-1.52_	0.66	_-1.66_	0.55	-1.51	0.47	_0.365_
GRF	mean HBF (BW)	PFC	-0.55	0.12	_-0.60_	0.13	-0.54	0.14	_0.026*_
GRF	peak VBF (BW)	FFC	2.42	0.59	2.55	0.55	2.70	0.52	_0.036*_
GRF	mean VBF (BW)	FFC	1.52	0.22	1.59	0.20	1.73	0.23	_<0.001**_
GRF	peak HBF (BW)	FFC	_-1.38_	0.38	-1.41	0.40	_-1.47_	0.29	_0.323_
GRF	mean HBF (BW)	FFC	-0.78	0.16	-0.84	0.17	_-0.95_	0.15	_<0.001**_
GRF	peak HBF ratio	both	0.89	0.54	0.93	0.39	1.09	0.53	_0.050*_
GRF	mean HBF ratio	both	1.46	0.38	1.44	0.38	1.94	0.78	_<0.001**_
Joint kinematics	peak KFA (°)	FFC	66.4	10.9	62.1	8.5	61.9	8.8	0.022*
Joint kinematics	KFA - IC (°)	FFC	21.9	6.3	23.5	6.1	23.8	6.7	0.264
Joint kinematics	KFA ROM (°)	FFC	_44.5_	8.4	38.6	9.0	38.0	7.9	_0.001**_
Joint kinematics	peak KAA (°) (- abduction, + adduction)	FFC	_-6.5_	7.5	-11.9	7.1	-14.7	7.5	_<0.001**_
Joint kinematics	KAA - IC (°) (- abduction, + adduction)	FFC	4.2	5.4	2.8	5.2	-0.2	5.4	<0.001**
Joint kinematics	KAA ROM (°)	FFC	_-10.7_	5.8	_-14.7_	5.1	-14.5	6.3	_<0.001**_
Joint kinematics	KRA - IC (°) (- internal, + external)	FFC	-9.9	9.3	-3.3	9.5	-0.8	9.8	_<0.001**_
Joint kinematics	peak KRA (°) (- negative, + external)	FFC	-11.2	8.7	_-4.9_	8.3	-4.8	7.2	_<0.001**_
Joint kinematics	HRA - IC (°) (- internal, + external)	FFC	11.0	8.5	9.7	9.3	9.4	11.0	0.684
Technique	Trunk inclination angle - IC (°) (relative to vertical line, + forward, - backward)	PFC	_6.9_	5.2	8.2	4.3	_8.4_	4.8	_0.117_
Technique	Trunk inclination angle - IC (°) (relative to vertical line, + forward, - backward)	FFC	_9.1_	8.1	_11.6_	6.7	_11.3_	_7.7_	_0.047*_
Technique	IFPA - IC (°) (+ internal, - external)	FFC	_9.4_	11.6	_22.2_	16.4	26.3	11.5	_<0.001**_
Technique	Lateral trunk flexion - IC (°) (- over stance leg, + direction of travel)	FFC	_-15.6_	6.5	_-21.4_	22.5	-18.6	7.6	_<0.001**_
Technique	Lateral foot plant distance - IC (m)	FFC	-0.29	0.04	-0.32	0.05	-0.34	0.05	<0.001**
Joint moment	peak KFM (Nm/kg)	FFC	3.07	0.65	3.30	0.64	3.64	0.77	<0.001**
Joint moment	peak KRM (Nm/kg) (- internal, + external)	FFC	_-0.62_	0.40	_-0.82_	0.52	_-1.16_	0.72	_<0.001**_
Joint moment	peak KAM (Nm/kg) (+ abduction, - adduction)	FFC	0.62	0.26	_0.98_	0.41	1.39	0.40	_<0.001**_

Key: VBF: Vertical braking force; HBF: Horizontal braking force; FFC: Final foot contact; PFC: Penultimate foot contact; IC: Initial contact; BW: Body weight; KFA: Knee flexion angle; ROM: Range of motion; KAA: Knee abduction angle; KRA: Knee rotation angle; IFPA: Initial foot progression angle; KFM: Knee flexion moment; KRM: Knee rotation moment; KAM: Knee abduction moment; ES: Effect size; CMAS: Cutting movement assessment scores; HRA: Hip rotation angle; *: $p \leq 0.05$; **: $p \leq 0.001$. Note: Italic denotes non-parametric

Table 6.5. Pairwise comparisons of 3D cutting mechanics between low, moderate, and high CMAS trials containing *p* values and effect size

	Variable	Foot contact	Low vs. Moderate				Low vs. High				Moderate vs. High			
			p	*g*	95% CI		*p*	*g*	95% CI		*p*	*g*	95% CI	
					LB	UB			LB	UB			LB	UB
GRF	**peak VBF (BW)**	**PFC**	0.426	-0.17	-0.52	0.18	0.122	-0.23	-0.63	0.17	0.293	-0.06	-0.38	0.25
	mean VBF (BW)	**PFC**	0.243	-0.28	-0.64	0.07	0.990	-0.09	-0.48	0.31	0.204	0.17	-0.14	0.49
	peak HBF (BW)	**PFC**	0.267	0.23	-0.12	0.58	0.236	-0.02	-0.42	0.37	0.845	-0.30	-0.62	0.02
	mean HBF (BW)	**PFC**	0.176	0.40	0.04	0.75	1.000	-0.11	-0.51	0.29	**0.045***	-0.48	-0.80	-0.15
	peak VBF (BW)	**FFC**	0.593	-0.22	-0.57	0.13	**0.034***	-0.50	-0.90	-0.10	0.259	-0.28	-0.60	0.04
	mean VBF (BW)	**FFC**	0.220	-0.33	-0.68	0.03	**<0.001****	-0.92	-1.33	-0.50	**<0.001****	-0.64	-0.97	-0.31
	peak HBF (BW)	**FFC**	0.705	0.08	-0.27	0.43	0.148	0.27	-0.13	0.67	0.235	0.17	-0.15	0.49
	mean HBF (BW)	**FFC**	0.093	0.37	0.02	0.73	**0.002***	1.08	0.66	1.50	**<0.001****	0.66	0.34	0.99
	peak HBF ratio	**both**	1.000	-0.08	-0.43	0.27	0.269	-0.38	-0.78	0.02	0.051	-0.35	-0.67	-0.03
	mean HBF ratio	**both**	0.927	0.04	-0.31	0.39	**<0.001****	-0.78	-1.18	-0.36	**<0.001****	-0.80	-1.13	-0.47
Joint kinematics	**peak KFA (°)**	**FFC**	0.032	0.43	0.08	0.79	**0.041***	0.45	0.05	0.85	1.000	0.03	-0.29	0.35
	KFA - IC (°)	**FFC**	0.461	-0.26	-0.61	0.09	0.380	-0.30	-0.69	0.10	1.000	-0.05	-0.36	0.27
	KFA ROM (°)	**FFC**	**0.001****	0.67	0.31	1.03	**0.002***	0.78	0.37	1.19	1.000	0.07	-0.25	0.38
	peak KAA (°)	**FFC**	**<0.001****	0.73	0.37	1.09	**<0.001****	1.09	0.66	1.51	0.075	0.39	0.06	0.71
	KAA - IC (°)	**FFC**	0.510	0.25	-0.11	0.60	**<0.001****	0.80	0.39	1.21	**0.001****	0.57	0.25	0.90
	KAA ROM (°)	**FFC**	**<0.001****	0.74	0.38	1.10	**0.001****	0.62	0.22	1.03	1.000	-0.04	-0.36	0.28
	KRA - IC (°)	**FFC**	**0.001****	-0.70	-1.06	-0.34	**<0.001****	-0.94	-1.36	-0.52	0.779	-0.25	-0.57	0.07
	peak KRA (°)	**FFC**	**0.001****	-0.74	-1.10	-0.38	**<0.001****	-0.81	-1.22	-0.40	1.000	-0.02	-0.34	0.29
	HRA - IC (°)	**FFC**	1.000	0.14	-0.21	0.50	1.000	0.16	-0.23	0.56	1.000	0.03	-0.29	0.35
Technique	**Trunk inclination angle - IC (°)**	**PFC**	0.051	-0.27	-0.63	0.08	0.073	-0.30	-0.70	0.10	0.974	-0.04	-0.36	0.28
	Trunk inclination angle - IC (°)	**FFC**	**0.043***	-0.33	-0.69	0.02	0.180	-0.28	-0.68	0.12	1.000	0.04	-0.28	0.36
	IFPA - IC (°)	**FFC**	**<0.001****	-0.90	-1.27	-0.54	**<0.001****	-1.45	-1.89	-1.00	0.064	-0.28	-0.60	0.03
	Lateral trunk flexion - IC (°)	**FFC**	**<0.001****	0.34	-0.01	0.70	**<0.001****	0.42	0.02	0.82	1.000	-0.16	-0.48	0.16
	Lateral foot plant distance - IC (m)	**FFC**	**0.025***	0.51	0.16	0.87	**<0.001****	1.06	0.64	1.48	**0.012***	0.45	0.12	0.77
Joint moment	**peak KFM (Nm/kg)**	**FFC**	0.188	-0.35	-0.71	0.00	**<0.001****	-0.79	-1.19	-0.37	**0.007***	-0.47	-0.79	-0.15
	peak KRM (Nm/kg)	**FFC**	**0.008***	0.43	0.08	0.78	**<0.001****	0.93	0.51	1.34	**0.002***	0.54	0.22	0.86
	peak KAM (Nm/kg)	**FFC**	**<0.001****	-1.07	-1.44	-0.70	**<0.001****	-2.25	-2.75	-1.74	**<0.001****	-0.98	-1.32	-0.65

Key: VBF: Vertical braking force; HBF: Horizontal braking force; FFC: Final foot contact; PFC: Penultimate foot contact; IC: Initial contact; BW: Body weight; KFA: Knee flexion angle; ROM: Range of motion; KAA: Knee abduction angle; KRA: Knee rotation angle; IFPA: Initial foot progression angle; KFM: Knee flexor moment; KRM: Knee rotation moment; KAM: Knee abduction moment; ES: Effect size; CMAS: Cutting movement assessment scores; CI: Confidence interval; LB: Lower bound; UB: Upper bound; HRA: Hip rotation angle; *: $p \leq 0.05$; **: $p \leq 0.001$; italic denotes non-parametric.

Trivial ES (≤ 0.19)	**Small ES (0.20-0.59)**	**Moderate ES (0.60-1.19)**	**Large ES (1.20-1.99)**	**Very Large ES (2.00-3.99)**	**Extremely large ES (≥4.00)**

6.5 DISCUSSION

The primary aim of this study was to examine the validity and relationship between the CMAS attained from a qualitative screening tool and peak KAM, quantified via 3D motion and GRF analysis. This study expanded on the preliminary work of Jones et al. (278) by using an additional camera filming at a higher sampling rate, and also investigating a larger sample size. In line with the study hypotheses, and substantiating Jones et al. (278), a very large (ρ = 0.796, p < 0.001) (Figure 6.2) relationship was observed between CMAS and peak KAM for participant average data, while large to very large relationships were observed for pooled data (Figure 6.3). Moreover, "higher-risk" cutting mechanics associated with greater knee joint loading, thus ACL injury risk, were displayed by participants and trials with higher CMASs (~7) compared to participants and trials with lower CMASs (~3) (Tables 6.3 & 6.5). The CMAS also demonstrated excellent intra-rater reliability (Table 6.2), and generally moderate to excellent inter-rater reliability (Table 6.2). Therefore, these findings indicate that the CMAS qualitative screening tool can be considered a reliable and valid method to identify athletes who generate high KAMs and "high-risk" cutting mechanics. This tool offers practitioners a field-based screening method which can be included in testing and screening batteries for cutting sports, so "high-risk" cutting deficits can be identified and "injury risk profiles" can be created for athletes.

In light of the "high-risk" cutting deficits associated with greater knee joint loads during side-step cutting (Appendix 5.1), Jones et al. (278) developed the CMAS screening tool and reported a large relationship between CMAS and peak KAM (ρ = 0.633; p < 0.001). Expanding on the preliminary investigation by Jones et al. (278), the present study observed a stronger relationship between CMAS and peak KAMs (ρ = 0.796, p < 0.001, Figure 6.2), in a substantially greater sample size (41 vs. 8 participants, 214 vs. 36 trials). The stronger relationships observed in the present study, compared to Jones et al. (278), could be attributed to the additional camera placed at 45° and increased sampling rate of the cameras (100 vs. 30 Hz). These additions may have permitted more accurate screening and evaluations of frontal and transverse plane deficits, such as trunk positioning and knee valgus. Nevertheless, these findings confirm that the CMAS is able to identify athletes who generate high peak KAMs, which offers practitioners a cheaper, time-efficient, and field-based applicable screening tool compared to 3D motion analysis, using only three high-speed cameras and free video analysis software.

While screening tools such as the LESS (430, 434), TJA (238, 397), and QASLS (8, 237) are useful for identifying abnormal and "high-risk" jump-landing mechanics, there is mixed evidence as to whether the examination of landing mechanics can identify athletes with poor cutting mechanics (4, 86, 302, 421). This issue is pertinent for practitioners who work with athletes who participate in cutting dominant sports. In addition, the LESS is the only screening tool to have been validated and assessed against 3D motion analysis (427, 434), with no evidence to suggest that the TJA and QASLS is capable of identifying athletes who generate high multiplanar knee joint loads. Conversely, in the present study, "higher-risk" cutting mechanics and greater multiplanar knee joint loads (Table 6.3) were demonstrated by participants and trials with high CMASs compared to participants and trials with low CMASs. These "higher-risk" mechanics included greater mean VBFs and HBFs, greater KAAs, greater lateral foot plant distances, greater IFPAs, and lower knee flexion ROM (Table 6.3), with moderate to large effect sizes. Moreover, greater multiplanar knee joint loads (KFMs, KAMs, KIRMs) were also demonstrated by participants and trials with high CMASs compared to low, with moderate to very large effect sizes (Tables 6.2 & 6.5). This finding is important because combined multiplanar loads strain the ACL to a greater extent compared to uniplanar loading (26, 342, 513). Krosshaug et al. (309) has highlighted the potential difficulties in estimating 3D joint kinematics based on 2D video evaluations of cutting mechanics. Conversely, the results indicate that the raters in the present study were capable of accurately evaluating and identifying aberrant lower-limb and trunk postures during cutting, as confirmed by the measurable differences in 3D kinetics and kinematics between participants and trials with "high" and "low" CMASs related to the CMAS scoring system (Tables 6.3 & 6.5).

Supporting the results of Jones et al. (278), higher CMASs were associated with greater peak KAMs (Figure 6.2), and "higher-risk" cutting mechanics were displayed by participants and trials with high CMASs (Table 6.3). These findings indicate that higher scores are representative of, in general, poorer cutting technique. The CMAS tool can therefore be useful for practitioners who want to screen and evaluate cutting movement quality to identify potentially "high-risk" athletes (240, 243, 354, 372), so these athletes can be targeted with biomechanical and neuromuscular informed training interventions to reduce potential injury risk (240, 243, 372). Qualitative screening tools such as the TJA (295, 540), LESS (135-137, 423, 438), and QASLS (120, 433) have been used to monitor the effectiveness of training interventions on jump-landing or single-leg control mechanics; therefore, the CMAS could be used to monitor pre-to-post changes in cutting movement quality in response to training interventions, and is subsequently a recommended future direction of research. However, it is emphasised that lower CMASs do not necessarily equate to optimal or "safe" technique, and practitioners should not only focus on total score, but focus on the CMAS criteria where athletes scored deficits (187, 278). For example, an athlete who scores 2-3 points may still display "high-risk" cutting deficits such as knee valgus, lateral trunk flexion, limited knee flexion, or hip internal rotation and thus, would still warrant specific injury risk mitigation training and conditioning. As such, practitioners should be conscious and are advised to look beyond the total CMAS score and use the CMAS tool to assist in the identification of potentially "high-risk" cutting deficits. The information attained from the CMAS may help inform the future prescription of training and conditioning to correct these deficits, and mitigate potential injury risk (240, 243, 372).

Although a plethora of investigations have focused on COD biomechanics associated with increased risk of injury and have identified a range of factors linked to greater knee joint loading (Table 6.1) (127, 189, 228, 267, 272, 273, 301, 358, 516, 519, 593), technical guidelines for coaching safer side-step cutting are limited. A unique aspect of the CMAS is that the criteria (Table 6.1) can be used as a technical framework for coaching safer side-step cutting which practitioners can use when working with their athletes (278). COD technique modification has been shown to be an effective modality for reducing "high-risk" mechanics and knee joint loading during COD (76, 115, 126). Consequently, using the CMAS as a screening tool and a technical framework for safer cutting could be a viable strategy which coaches and practitioners could use to identify specific "high-risk" cutting deficits (i.e., lateral trunk flexion, knee valgus etc) to help inform COD technique modification training.

It is worth noting, however, that some of the "high-risk" cutting deficits may be required for faster cutting performance (115, 148, 183, 228). For example, a wide lateral foot plant is necessary to generate MLPF and impulse (228, 272), thus subsequent exit velocity; however, this technique concurrently elevates peak KAMs (127, 228, 272, 301). Limited knee flexion and motion is associated with shorter GCTs (115, 183) (Chapter 5), but this posture increases KAMs (301, 593), KFMs and GRFs (115, 640), thus potential ACL loading (183). Moreover, lateral trunk flexion, from an attacking and evasive perspective, may be performed to feint and deceive opponents (61, 62, 252), but is a critical factor that augments potentially hazardous KAMs (245, 248, 366). Consequently, practitioners should acknowledge the trade-off between knee joint loading (injury risk) and performance when screening cutting mechanics, because some of the "high-risk" deficits demonstrated could be effective for performance. Nonetheless, practitioners should ensure that their athletes' have the physical capacity (i.e., neuromuscular control, co-contraction, and rapid force production) to tolerate the knee joint loading demands of side-steps (272, 328, 341, 432). Further research is required to improve our understanding of the potential "performance-injury conflict" during cutting (183).

6.6 LIMITATIONS

It should be acknowledged that, due to the multiplanar nature of side-step cutting (39), some athletes pre-rotate towards the direction of travel during WA of the cut (516). This pre-rotation can potentially result in parallax error because the athlete is not perpendicular to the cameras which can restrict evaluations of particular CMAS criteria using the frontal plane and 45° cameras. Additionally, the current study only investigated a side-step cutting action; thus, the CMAS screening tool is specific to side-step cutting only. Specific screening tools must be developed and validated for assessing other

COD actions, such as XOCs and pivots, which are also performed and associated with injury in multidirectional sport (94, 270). However, side-step cutting appears to be the predominant COD action associated with non-contact ACL injury (94, 376); therefore, this highlights the importance and inclusion of side-step cutting screening tools (CMAS) in testing batteries for athletes who participate in cutting sports, such as soccer, rugby, handball, American football, and badminton. Furthermore, the intra- and inter-rater reliability, generally, was moderate to excellent (Table 6.2), but limited to biomechanists and strength and conditioning coaches. Further work is required to establish agreements and reliability between different applied practitioners, such as sports rehabilitators, physiotherapists, and sports coaches, in order to confirm its efficacy in the field. Finally, a pre-planned cutting task was used in the present study; however, results of previous research have shown that unplanned side-stepping results in greater knee joint loads, more abnormal mechanics, and less muscle support to counteract the greater loads compared to pre-planned side-stepping (37, 38, 65).

6.7 CONCLUSION

In conclusion, large to very large significant relationships were observed between CMAS and peak KAM, and "higher-risk" cutting mechanics associated with greater knee joint loading were demonstrated by participants and trials with "high" CMASs (~7) compared to participants and trials with "low" CMASs (~3). As such, the CMAS is a valid and reliable screening tool for evaluating side-step cutting movement quality and offers practitioners a cost-effective and easily applicable field-based screening tool to identify athletes who generate high peak KAMs during side-step cutting. Practitioners should therefore consider including the CMAS in their fitness and testing batteries when screening and profiling athletes who participate in multidirectional sports. Equally, the CMAS allows practitioners to identify "high-risk" cutting deficits in athletes and subsequently create an "injury risk profile". These identified deficits can be targeted and addressed through biomechanical and neuromuscular informed training interventions. Finally, the CMAS can be used as a potential technical framework for coaching "safer" cutting.

CHAPTER 7: The Effects of Six-Weeks Change of Direction Speed and Technique Modification Training on Cutting Performance and Movement Quality in Male Youth Soccer Players

7.1 INTRODUCTION

Soccer is classified as an intermittent sport, composed of maximal or near-maximal (effort) high-intensity actions, such as accelerations, sprints, CODs, decelerations, jumps, and kicks, which are then interspersed with brief recovery periods of low-intensity actions, such as walking and jogging (51, 539, 566). In particular, time-motion analysis has revealed soccer players, on average, perform 609 ± 193 cuts of 0° to 90° to the left or right (51) during a match, typically in response to an opponent, the ball, or to create space. Similarly, Robinson et al. (487) found 38.9 ± 13.3 and 36.3 ± 12.9 directional changes (45-135° movement 4 m/s or faster) were performed to the left and right, respectively, by soccer players during match play. Moreover, COD (≥ 50°) actions which are then followed by a sprint are associated with critical moments, such as goal scoring and assists in soccer (171). Consequently, given the frequency of COD actions performed in soccer, and its association with decisive moments (i.e., goal scoring), the ability to change direction rapidly can be considered an important quality to develop.

COD speed is defined as "the ability to decelerate, reverse, or change movement direction, and accelerate again" (271), and as previously stated, soccer players frequently perform rapid decelerations, directional changes, and sprints to create space, or to react to an opponent or ball. The determinants of COD speed are multifaceted (410, 510, 628, 629), and influenced by a range of physical lower-limb qualities (strength capacity, rate of force development, power, reactive strength), linear speed, and technical factors (body lean and posture, foot placement, adjustment of strides) (409, 410, 510, 628, 629). Enhancements in COD speed have been demonstrated as a consequence of lower-limb strength training (56, 67, 194, 284, 414, 544, 587) and plyometric training (13, 15, 194, 352); however, COD speed training interventions over 6-12 weeks (i.e., field-based sprint, deceleration, and COD drills) are also effective in improving COD speed performance in soccer players (77, 81, 87, 347, 348, 460) and college athletes (332, 630). Moreover, a recent meta-analysis has confirmed that COD speed and sprint training interventions elicit short-term improvements in COD speed performance in soccer players (194), while augmentations in bilateral and unilateral jump height and distance, reactive strength, and eccentric and concentric knee flexor and extensor strength have been observed following COD speed training interventions (81, 87, 281, 332, 460, 532). As such, COD speed training provides practitioners with a relatively easy to perform field-based method to enhance COD performance and other physical qualities with the use of minimal equipment. These aforementioned advantages associated with COD speed training, rationalise and highlight its importance and inclusion in soccer players strength and conditioning programmes (477, 566).

While cutting is important for successful performance in soccer, it should be noted that directional changes are also key actions associated with non-contact ACL injury in this sport (63, 204, 583). This occurrence can be attributed to the propensity to generate large multiplanar knee joint loading, such as KAMs and KIRMs (39, 126, 127, 273, 301) during the plant foot contact, which increases ACL strain (26, 288, 342, 424, 513). When these knee joint loads are high enough, it can result in mechanical failure and ACL rupture (327, 365). ACL injury risk factors are multifactorial (i.e., hormonal, anatomical, biomechanical, neuromuscular, environmental) (241, 469) and are influenced by both internal and external factors (48, 365, 469). Researchers have identified a number of key biomechanical and neuromuscular control deficits. These include the following: wide lateral foot plant distances (127, 228, 272, 301), greater hip abduction angles (519, 593), increased IFPAs (274, 519), increased initial hip internal rotation angles (228, 358, 516, 519), greater initial and peak KAA (272, 274, 301, 358, 516), reduced knee flexion (76, 593), and greater lateral trunk flexion over the plant leg (127, 141, 189, 267, 272). These deficits have been identified as "high-risk" postures during directional changes, and are thus associated with greater knee joint loading (148, 183) and a heightened ACL injury risk (242, 298, 342, 343, 512, 614).

Additionally, the aforementioned "high-risk" biomechanical and neuromuscular control deficits have been identified as potential injury risk factors in soccer (475, 479), and while high KAMs are associated with ACL loading (39, 126, 127, 273, 301), high KAMs are also associated with patellofemoral pain (PFP) (393, 395, 580); a common knee injury in soccer (217, 404, 422). Moreover, with the exception of lateral foot plant distance (228, 261, 272) and limited knee flexion (115), the "high-risk" COD postures offer no associated performance benefits (183, 228, 346), and in fact, reducing lateral trunk flexion and encouraging a trunk lean towards the direction of travel could be a faster technique (183, 228, 346). Importantly, however, biomechanical and neuromuscular control deficits are modifiable (241, 432, 435, 475, 479), with previous studies showing that COD technique modification training (i.e., coaching cues and feedback to reduce postures associated with increased knee joint loads) can reduce hazardous knee joint loading in athletes (76, 115, 126, 277, 292). Therefore, addressing biomechanical and neuromuscular control deficits could be a viable strategy to reduce injury risk in soccer players (148, 183, 432, 435, 436, 479); however, no study, to date, has examined the effect of COD technique modification training on cutting movement quality in soccer players.

In soccer, it has been reported that ACL injury rates can range from 0.06 to 3.7 per 1000 hours (training/match exposures) (49, 173, 582), yet despite the improvements in sports medicine and strength and conditioning practices in more recent times, non-contact ACL injuries are not declining and still remain an issue (63, 204, 582, 583). ACL injuries can be career-threatening (313), with a plethora of negative economic (112, 241, 468), psychological (241, 315), and health (241, 333) consequences. Despite a high return to play rate following an ACL injury in soccer within a year of injury (≥ 90%), only two-thirds of players play at the same competitive level three years later (582). As such, ACL injury mitigation strategies are of vital importance in soccer players. One strategy to help reduce ACL injury risk in soccer players is evaluating movement quality to identify athletes that display abnormal and "high-risk" mechanics (185, 241, 478, 479). This information can then be used to inform the future prescription of training and conditioning, so specific deficits can be targeted through appropriate training interventions to decrease the relative risk of injury (243, 354, 372, 436). 3D motion analysis is considered the gold standard (187, 241) for evaluating movement mechanics, and has been previously been used to monitor changes in COD mechanics (76, 115, 126, 277, 292); however, this is a complex, time-consuming, and expensive process, which is often restricted to laboratory settings, thus limiting its application for practitioners working in field-based settings (163, 187, 241, 302, 408, 527).

Recently, the CMAS has been created and validated against 3D motion analysis (278)(Chapter 6), with strong relationships observed between CMAS and peak KAMs (ρ = 0.633-0.796; p < 0.001). As practitioners will implement training interventions to reduce "high-risk" cutting mechanics, it is imperative that the effectiveness of such interventions can be monitored using a valid and reliable screening tool. The CMAS provides practitioners with a valid field-based screening tool to identify athletes who generate high KAMs and poor movement quality; however, to the best of the author's knowledge, no study has monitored the effectiveness of training interventions on cutting movement quality using a field-based screening tool, with previously published training interventions monitoring changes in the neuromuscular and biomechanical deficits using the TJA (295, 540), LESS (135-137, 423, 438), and QASLS (120, 433).

Given the importance of cutting ability for performance in soccer players (51, 171, 487), and its association with ACL injury (63, 204, 583), it is imperative that soccer players adopt training strategies that help improve performance while reducing injury risk. Improvements in COD speed have been demonstrated following COD speed interventions (77, 81, 87, 347, 348, 460), while reductions in COD knee joint loading have been observed following technique modification (76, 115, 126, 277). However, it is somewhat surprising that the effect of COD speed and technique modification training on performance and injury risk factors in male soccer players has yet to be established. Therefore, the aim of this study was to determine the effects of a six-week COD speed and technique modification training intervention on cutting performance and movement quality in male youth (U17-U18s) soccer players using the field-based CMAS screening tool and timing gates to assess COD quality and performance, respectively. Since the introduction of the elite performance player plan, injury rates in adolescents have increased three-fold (481), and because youth players are striving for professional contracts, injury mitigation is of great importance for this population (481, 482). If athletes can reduce "high-risk" postures (i.e., reduce CMAS score) and improve performance (i.e., completion time and COD deficit), the COD speed and technique intervention can be considered successful and could be adopted by other practitioners. However, if athletes can reduce "high-risk" postures while maintaining cutting performance; it can still be viewed as a positive effect. This is important because athletes are driven by performance, and are less to likely to adopt safer strategies at the expense of faster, effective performance (183, 228). It was hypothesised that a COD speed and technique modification intervention would result in faster cutting performance and improved cutting movement quality in comparison to a CG.

7.2 METHODS

7.2.1 Research design

A non-randomized, controlled 6-week intervention study with a repeated measures pre-to-post design was used. Youth soccer players (U17s) from an English professional soccer club were recruited for the IG which consisted of a 6-week COD speed and technique modification training programme (Appendix 6.1), consisting of two 20-minute sessions per week. These sessions replaced the soccer teams' normal warm-ups for two of the sessions. Conversely, youth soccer players from the same club (U18s) acted as the CG and continued their normal field-based warm-ups (mobilisation, low-level jump-landing and sprint drills). Pre-to-post assessments of 70˚ cutting performance, 10-m sprint times, and COD deficit was assessed to monitor the effectiveness of the training intervention, while cutting movement quality was assessed with the recently developed and validated CMAS screening tool (278) (Chapter 6).

7.2.2 Participants

26 male youth soccer players from an English professional soccer club (youth-teams, at the time of study, played in a regional league against youth teams of a similar standard, participated in the FA youth cup, and represented Manchester county; first team played in the 5th tier in English football league at the time of the study) were recruited and participated in this study. Based on an effect size of 1.15 for pre-to-post changes (dependent T-Test) in COD speed performance in youth soccer players following COD speed training (77), *a priori* analysis, using G*Power (Version 3.1, University of Dusseldorf, Germany) (172), indicated a minimum sample size of 8 was required to achieve a power of 0.80, and type 1 error or alpha level of 0.05. Thirteen soccer players from the U17s squad (consisting of defenders, midfielders, and attackers) (age: 16.9 ± 0.2 years; height: 1.77 ± 0.05 m; mass: 69.2 ± 9.2 kg) were recruited for the IG. Conversely, thirteen soccer players from the U18s squad (age: 17.8 ± 0.3 years;

height: 1.77 ± 0.07 m; mass: 73.3 ± 8.1 kg) acted as the CG and continued their normal field-based warm-ups (i.e., 5 minutes mobilisation exercises before progressing to 15 minutes of low-level jump-landing and sprint drills). These sample sizes were in line with those used in previous COD speed training (77, 81, 87, 347, 348, 630) or COD technique modification studies (76, 126, 277). Goalkeepers were not included in this investigation (77). The investigation was approved by the Institutional Ethics Review Board (HSR1617-131), and all participants were informed of the benefits and risks of the investigation prior to signing an institutionally approved consent form to participate in the study.

To remain as an active participant in the study and used for further analysis, participants were not allowed to miss more than three of the 12 sessions in total (i.e., ≥ 75% compliance rate). Subsequently, due to match-related injuries or illness, five and two participants withdrew from the intervention and CGs, resulting in sample sizes of 8 and 11 (Figure 7.1), respectively. Participants in the IG completed on average 10.6 ± 1.2 sessions over the intervention (88.5 ± 9.9%), while the CG completed 10.8 ± 1.3 sessions (90.0 ± 10.8%) over the intervention period.

All soccer players from both groups possessed at least five years' soccer training experience and had never sustained a knee injury such as an ACL injury prior to testing. All participants participated in the same technical and tactical soccer sessions (led by head soccer coach), five times a week (~90 minutes sessions on match day +2 days, +3 days, -3 days, -2 days, -1 day). Additionally, all participants performed two strength training sessions (session 1: match day +2 days, session 2; match day -3 days) per week and received the same training programmes (i.e., exercises, sets, reps, intensity). At the time of the training intervention, all players were in a strength mesocycle and played one competitive match a week. All of the procedures were carried out during the competitive season to ensure that no large physical changes were made as a result of the conditioning state (81).

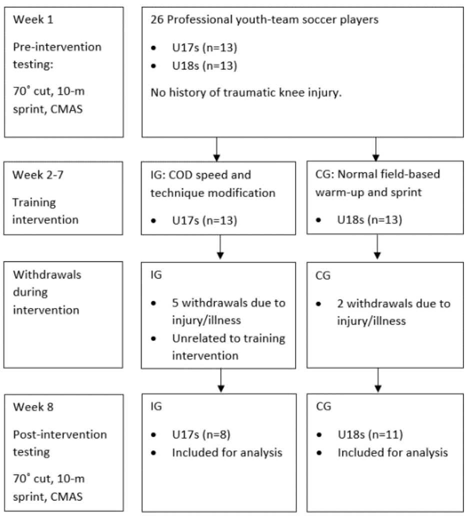


Figure 7.1. Flow diagram of participant participation throughout all stages of the intervention study. IG: Intervention group; CG: Control group; CMAS: Cutting movement assessment score

7.2.3 Procedures and field tests

All pre-to-post testing (week 1 and week 8) was performed on the same weekday (Figure 7.1), and at the same time of day (10:00-12:00) to control for circadian rhythm (154), which coincided with normal skills training time. Due to facility constraints, field-testing was performed in an indoor sports hall on a hardwood court. All participants rested the day before testing and were asked to attend testing in a fed and hydrated state, similar to their normal practices before training. On arrival, all participants had their BM (Seca Digital Scales, Model 707, Birmingham, United Kingdom) and standing height (Stadiometer; Seca, Birmingham, United Kingdom) measured to the nearest 0.1 kg and 0.1 cm, respectively. All participants then performed a standardised 10-minute warm up led by the principal investigator which consisted of dynamic stretches, low-level plyometrics, and progressive sprints drills, before completing the cutting and 10-m sprint assessments, similar to previous research (150).

7.2.3.1 70˚ cutting task

The cutting task consisted of a 5-m approach, 70˚ cut, and 5-m exit towards the finish (Figure 7.2). Completion time was measured using sets of single beam Brower photocell timing gates (Draper, UT, USA) setup at the start and finish (Figure 7.2); time was recorded to the nearest 0.001 seconds. Timing gates were placed at the approximate hip height for all athletes as previously recommended (623), to ensure that only one body part (such as the lower torso) breaks the beam. Participants started 0.5-m behind the first gate, to prevent any early triggering of the initial start gate, from a two-point staggered start, and were instructed to sprint as fast as possible, cut (lateral foot plant), and reaccelerate as fast as possible through the finish (Figure 7.2). Participants performed four practice trials at 75% of maximum perceived effort, before performing six maximum effort cutting trials: three changing direction with a left foot plant, and three changing direction with a right foot plant, interspersed with two-minutes' rest between trials. The testing order was counterbalanced. If the participants' foot did not touch the cutting line (i.e., cut prematurely), slipped, turned off the incorrect foot, or did not perform a side-step cutting action, the trial was disregarded, and another trial was performed following the rest period. The mean of three trials from each limb was used for further analysis (3, 412).

To permit qualitative screening of cutting technique, three Panasonic Lumix FZ-200 high-speed cameras sampling at 100 Hz filmed the cutting trials. These cameras were positioned on tripods 3.5-m away from the cutting point at a height of 0.60-m and were placed in the sagittal and frontal plane, with a camera also placed 45˚ relative to the cut in accordance with the recommendations by Jones et al. (278) (Figure 7.2).

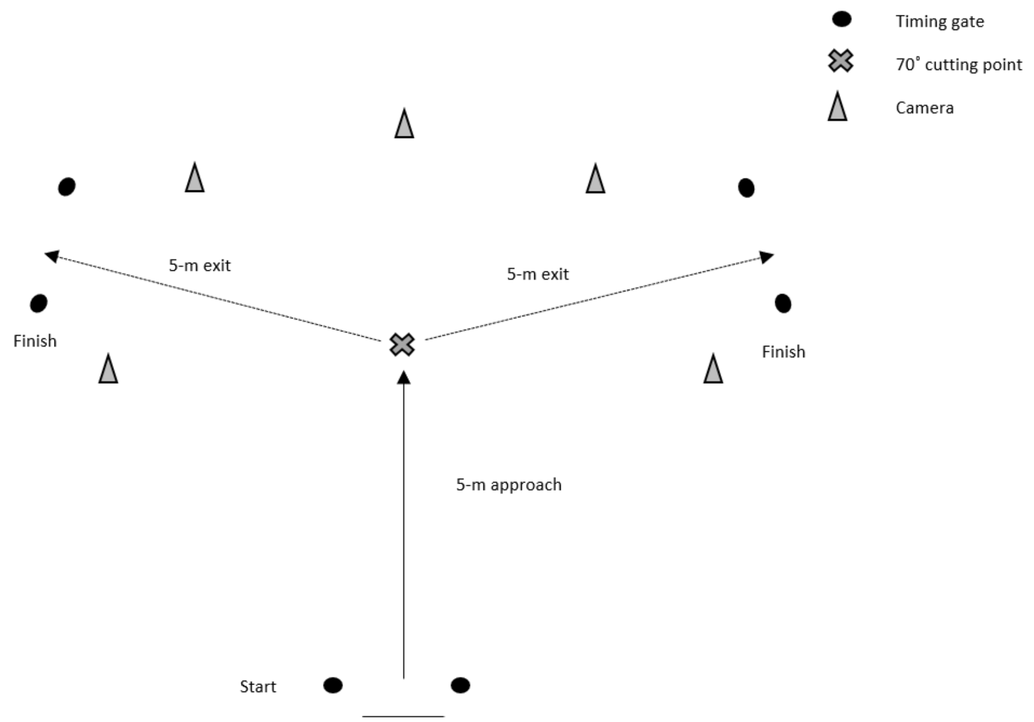


Figure 7.2. Schematic representation of the 70˚ cutting task

<u>7.2.3.2 10-m sprint</u>

Following the cutting assessment, participants were provided with five minutes' rest before completing the sprint assessment. Two 10-m sprint warm-up trials at 50% and 75% of maximum perceived effort were given for all participants. All participants performed three maximum effort sprint trials, with two minutes' rest between trials, using the same timing gates as described above placed at 0- and 10-m. Participants started 0.5-m behind the first gate, to prevent any early triggering of the initial start gate, from a two-point staggered start (330). The mean of three trials was used for further analysis (3).

<u>7.2.3.3 COD deficit</u>

To provide a more isolated measure of COD ability (144, 411-413), COD deficit was calculated using the formula: mean 70˚ cutting completion time – mean 10-m sprint time (113, 412, 413, 494). COD deficit was calculated for left and right cutting performances.

<u>7.2.3.4 CMAS screening</u>

Table 6.1 presents the CMAS qualitative screening analysis tool to estimate the magnitude of peak KAMs during cutting which has been validated against 3D motion analysis (278) (Chapter 6). If an athlete exhibits any of the characteristics/deficits in Table 6.1, they were awarded a score, with a higher score representative of poorer technique and potentially greater peak KAMs (278) (Chapter 6).

All video footage was viewed in Kinovea software (0.8.15 for Windows, Bordeaux, France), which is free, and was used for qualitative screening using the CMAS (Table 6.1). This software allowed videos to be played at various speeds and frame-by-frame. All videos were screened within two-weeks pre- and post-testing. Two raters screened the videos: the lead researcher who possesses seven years' strength and conditioning and biomechanics experience, viewed and graded each trial; the second rater was a graduate in Strength and Conditioning and possessed two years' strength and conditioning experience. The second rater viewed and screened one trial from each participant and these scores were compared to the lead researcher to establish inter-rater reliability. Raters were allowed to independently watch the videos as many times as necessary (182, 427), at whatever speeds they needed to score each test, and could also pause footage for evaluative purposes (182). On average, qualitative screening of one cutting trial took ~3 minutes. Prior to qualitative screening, all raters attended a one-hour training session outlining how to grade the cutting trials using the CMAS, and to establish and uniformly agree on "low-risk" and "high-risk" movement patterns using pilot video footage (Appendix 6.2)

<u>7.2.3.5 6-week COD speed and technique modification training intervention</u>

A six-week COD speed and technique modification intervention, described in Appendix 6.1, was performed by the IG twice a week (20 minutes per session) (session 1: match day +3 days; session 2: match day -2 days), with minimum 48 hours between sessions. The six-week technique modification intervention focused on pre-planned low intensity decelerations and turns (weeks 1-2), before progressing intensity via velocity and angle (weeks 3-4) (56, 131, 148, 411), and introducing a stimulus with increased intensity (weeks 3-6). The COD programme was in accordance with COD speed recommendations from the NSCA (215), Nimphius (410), and recent braking strategy recommendations (147), and the duration, distances, and number of CODs were similar to previously successful 6-week COD speed (77, 81, 87, 347, 348, 630) and COD technique modification studies (126, 277). The sessions were led by the lead researcher who is a certified strength and conditioning specialist with extensive experience in coaching COD speed and agility drills.

All COD speed and technique modifications sessions took place at the soccer team's training facility, with the first session of the week performed on a synthetic astro-turf and the second session of the week performed on a synthetic third generation rubber crumb field-turf. Players were given individual feedback regarding their technique, and because Chapter 5 (Study 2) confirmed that high MLPFs over short GCTs were associated with faster cutting performance, reduced lateral trunk flexion was associated with shorter GCTs, PFC dominant braking strategies were associated with faster cutting performance and lower KAMs, and greater KAAs offered no performance benefits and increased knee joint loads, external verbal coaching cues such as "slam on the brakes" (to promote PFC braking), "push/punch the ground away" (to enhance MLPF and encourage active knee flexion) (631), and "face towards the direction of travel" (to reduce lateral trunk flexion) were used to promote safer mechanics (34, 598), promote faster performance (35, 461, 610), and facilitate better retention (88, 169, 598, 620).

7.3 STATISTICAL ANALYSES

All statistical analyses were performed using SPSS v 25 (SPSS Inc., Chicago, IL, USA) and Microsoft Excel (version 2016, Microsoft Corp., Redmond, WA, USA). The primary outcome variables of this study were cutting completion time, COD deficit (CODD), 10-m sprint time, and CMAS.

7.3.1. Reliability – within session reliability of primary outcomes

Within-session reliability for the primary outcome variables was assessed for each group and session using ICCs (two-way mixed effects, average measures, absolute agreement), CV%, and SEM. The CV% was calculated as SD/mean × 100 for each participant and averaged across participants. The SEM was calculated using the formula: SD(pooled) * $\sqrt{(1 - \mathrm{ICC})}$ (554), whereas the SDD was calculated from the formula: (1.96 * ($\sqrt{(2)}$) * SEM (100, 306). ICCs were interpreted based on the following scale presented by Koo and Li (299): poor (< 0.50), moderate (0.50-0.75), good (0.75-0.90), and excellent (> 0.90). Minimum acceptable reliability was determined with an ICC >0.7 and CV <15% (28, 214).

7.4.2 CMAS inter- and intra-rater reliability

To determine intra-rater reliability, 26 trials (one from trial from each participant) were randomly selected by the lead researcher, and each trial was viewed and graded on two separate occasions separated by seven days, in line with previous research (182, 434). Similarly, for inter-rater reliability, 19 trials were screened by the other researcher and these scores were compared to the lead researcher. ICCs and SEMs for the CMAS total score were determined. For each item within the CMAS (Table 6.1), percentage agreements (agreements /agreements + disagreements × 100) and Kappa coefficients were calculated (278). Kappa coefficients were calculated using the formula: *k* = Pr(a) – Pr(e) / 1 – Pr(e), where Pr(a) = relative observed agreement between raters; Pr(e) = hypothetic probability of chance agreement, which describes the proportion of agreement between the two methods after any agreement by chance has been removed (95, 216, 577). The kappa coefficient was interpreted based on the following scale of Landis and Koch (314): slight (0.01-0.20), fair (0.21-0.40), moderate (0.41-0.60), good (0.61-0.80), and excellent (0.81-1.00). Percentage agreements were interpreted in line with previous research (107, 427), and the scale was as follows: excellent (>80%), moderate (51-79%), and poor (≤ 50%) (107, 427).

7.4.3 Pre-to-post changes in primary outcomes

Normality was inspected for all variables using a Shapiro-Wilks test. A two-way mixed ANOVA (group; time) with group as a between-subjects factor measured at 2 levels (IG and CG), and time (pre- and post-training measures) the within-subject factor. This was used to identify any significant main (time) or interaction (group × time) effects for primary outcome variables between IG and CG, pre-to-post testing. A Bonferroni-corrected pairwise comparison design was used to further analyse the effect of the group when a significant interaction effect was observed for time and group. Partial eta squared effect sizes were calculated for all ANOVAs with the values of 0.010-0.059, 0.060-0.149, and ≥ 0.150 considered as small, medium, and large, respectively, according to Cohen (96).

Furthermore, pre-to-post changes in primary outcome variables for each group were assessed using paired sample t-tests for parametric data and Wilcoxon-sign ranked tests for non-parametric data. Magnitudes of differences were assessed using Hedges' *g* effect sizes, mean change, and percentage change ((post-pre)/pre × 100) with 95% CIs. Hedges' *g* effect sizes were calculated as described previously (231) (Chapter 4.4) and interpreted as trivial (≤ 0.19), small (0.20 – 0.59), moderate (0.60 – 1.19), large (1.20 – 1.99), very large (2.00 – 3.99), and extremely large (≥ 4.00) (255). Comparisons in pre- and post-intervention primary outcome variables and change in primary outcome variables between IG and CGs were also assessed using independent sample t-tests or Mann-Whitney U tests, with effect sizes outlined above. Statistical significance was defined as $p \leq 0.05$ for all tests.

7.4 RESULTS

7.4.1. Within-session reliability measures

Within-session reliability for the IG and CG pre- and post-intervention primary outcome variables are presented in Tables 7.1 and 7.2 containing ICCs, CV%, SEM, and SDD. All variables for the IG displayed good to excellent ICCs pre- and post- intervention (Table 7.1). Cutting completion times and 10-m sprint times displayed low levels of variance pre- and post-intervention. CODDs and CMASs displayed high levels of variance pre- and post-intervention (Table 7.1). All variables for the CG displayed moderate to excellent ICCs pre- and post- intervention, excluding left cut completion time and CODD pre-intervention which displayed lower ICCs (Table 7.2). Cutting completion times and 10-m sprint times displayed low levels of variance pre- and post-intervention. CODDs and CMASs displayed high levels of variance pre- and post-intervention (Table 7.2).

Table 7.1. IG within-session reliability measures

IG pre-intervention within-session reliability

Variable		ICC	LB	UB	CV%	LB	UB	SEM	SDD	SDD%
Completion time	Right Cut (s)	0.944	0.800	0.988	2.8	1.8	3.7	0.042	0.117	5.0
	Left Cut (s)	0.865	0.537	0.971	2.8	1.4	4.2	0.054	0.150	6.4
CODD	Right CODD (s)	0.889	0.622	0.976	15.1	8.3	21.9	0.043	0.120	25.9
	Left CODD (s)	0.843	0.461	0.966	13.9	8.6	19.3	0.055	0.153	33.0
Sprint	10-m sprint (s)	0.861	0.561	0.969	2.1	0.9	3.4	0.034	0.093	5.0
CMAS	CMAS Right	0.934	0.776	0.986	11.4	5.2	17.6	0.49	1.36	21.7
	CMAS Left	0.865	0.526	0.971	15.7	9.3	22.2	0.67	1.84	29.3

IG post-intervention within-session reliability

Variable		ICC	LB	UB	CV%	LB	UB	SEM	SDD	SDD%
Completion time	Right Cut (s)	0.934	0.786	0.985	2.0	1.6	2.4	0.026	0.071	3.5
	Left Cut (s)	0.917	0.732	0.982	2.3	1.1	3.4	0.037	0.102	4.8
CODD	Right CODD (s)	0.873	0.596	0.972	20.8	15.0	26.6	0.027	0.074	34.7
	Left CODD (s)	0.870	0.580	0.972	17.0	10.5	23.4	0.038	0.106	38.4
Sprint	10-m sprint (s)	0.932	0.776	0.985	1.8	0.7	2.9	0.027	0.076	4.1
CMAS	CMAS Right	0.870	0.553	0.972	18.3	12.4	24.3	0.59	1.63	33.9
	CMAS Left	0.817	0.373	0.961	22.2	16.0	28.3	0.63	1.74	42.6

Key: IG: Intervention group; CODD: Change of direction deficit; CMAS: Cutting movement assessment score; ICC: Intraclass correlation coefficients; CV%: Coefficient of variation; LB: Lower bound 95% confidence interval; UB: Upper bound 95% confidence interval; SEM: Standard error of measurement; SDD: Smallest detectable difference

Table 7.2. CG within-session reliability measures

CG pre-intervention within-session reliability

Variable		ICC	LB	UB	CV%	LB	UB	SEM	SDD	SDD%
Completion time	Right Cut (s)	0.878	0.671	0.964	3.0	2.1	3.9	0.051	0.141	5.9
	Left Cut (s)	0.693	0.180	0.909	2.9	1.9	4.0	0.061	0.168	6.9
CODD	Right CODD (s)	0.847	0.585	0.955	14.9	9.6	20.3	0.052	0.145	28.3
	Left CODD (s)	0.661	0.096	0.899	13.2	8.7	17.7	0.062	0.172	31.5
Sprint	10-m sprint (s)	0.762	0.311	0.931	2.3	1.7	3.0	0.035	0.096	5.1
CMAS	CMAS Right	0.898	0.723	0.970	14.4	9.9	18.8	0.52	1.43	28.3
	CMAS Left	0.877	0.668	0.964	18.5	9.8	27.3	0.56	1.55	33.9

CG post-intervention within-session reliability

Variable		ICC	LB	UB	CV%	LB	UB	SEM	SDD	SDD%
Completion time	Right Cut (s)	0.727	0.214	0.921	3.1	2.2	4.1	0.059	0.163	7.2
	Left Cut (s)	0.903	0.738	0.972	2.0	1.3	2.7	0.033	0.092	4.0
CODD	Right CODD (s)	0.721	0.197	0.920	18.8	14.1	23.6	0.059	0.163	44.7
	Left CODD (s)	0.932	0.816	0.980	11.6	7.4	15.7	0.032	0.089	21.3
Sprint	10-m sprint (s)	0.830	0.546	0.950	2.4	1.4	3.3	0.036	0.099	5.3
CMAS	CMAS Right	0.854	0.609	0.957	19.2	14.1	24.2	0.63	1.76	33.9
	CMAS Left	0.774	0.355	0.935	15.3	8.9	21.7	0.65	1.81	36.0

Key: CG: Control group; CODD: Change of direction deficit; CMAS: Cutting movement assessment score; ICC: Intraclass correlation coefficients; CV%: Coefficient of variation; LB: Lower bound 95% confidence interval; UB: Upper bound 95% confidence interval; SEM: Standard error of measurement; SDD: Smallest detectable difference

7.4.2. CMAS inter- and intra-rater reliability

Table 7.3. Intra- and inter-rater reliability for CMAS criteria and total score

Variable/ CMAS tool criteria	Intra-rater reliability		Inter-rater reliability	
	% Agreement	*k*	% Agreement	*k*
Clear PFC braking	92.3	0.755	94.7	0.894
Wide lateral leg plant	96.2	0.920	100.0	1.000
Hip in an initial internally rotated position	100.0	1.000	100.0	1.000
Initial knee 'valgus' position	96.2	0.866	89.5	0.789
Inwardly rotated foot position	96.2	0.922	100.0	1.000
Frontal plane trunk position relative to intended direction	96.2	0.935	94.7	0.906
Trunk upright or leaning back throughout contact	96.2	0.906	100.0	1.000
Limited knee flexion during final contact	96.2	0.923	94.7	0.872
Excessive knee 'valgus' motion during contact	96.2	0.920	94.7	0.906
Average	96.2	0.905	96.5	0.930

Key: PFC: Penultimate foot contact; CMAS: Cutting movement assessment score

Excellent intra- (ICC = 0.972, SEM = 0.32, SDD = 0.88) and inter-rater reliability (ICC = 0.917, SEM = 0.38, SDD = 1.05) was observed for CMAS total score. Intra- and inter-rater percentage agreements and Kappa coefficients for intra-and inter-rater reliability are presented in Table 7.3. Excellent intra- (≥ 92.3%, $k \geq 0.866$) and inter-rater (≥ 89.5% $k \geq 0.872$) percentage-agreements and kappa coefficients were demonstrated for all CMAS variables (Table 7.3), except for PFC braking ($k = 0.755$) and initial knee valgus position ($k = 0.789$) which demonstrated good kappa coefficients.

7.4.3. Pre-to-post changes in primary outcomes

Pre-to-post changes in primary outcome variables for the IG and CG are presented in Tables 7.4 and 7.5, containing descriptives, p values, effect sizes, percentage differences, and mean differences. No significant differences were observed in primary outcome variables between groups pre-intervention ($p > 0.05$); however, effect sizes indicated that the IG displayed faster right cut ($p = 0.468$, $g = -0.33$), left cut ($p = 0.097$, $g = -0.78$), right CODD ($p = 0.432$, $g = -0.36$), left CODD ($p = 0.062$, $g = -0.77$) and greater right ($p = 0.171$, $g = 0.71$) and left CMASs ($p = 0.456$, $g = 1.09$) compared to the CG. Trivial and non-significant ($p = 0.844$, $g = -0.09$) differences were observed in 10-m sprint times between groups.

Table 7.4. IG pre-to-post changes

Variable	IG Pre		IG Post		Hedges' *g* effect size				% change				Mean difference				Change greater than pre SDD	
	Mean	**SD**	**Mean**	**SD**	***p***	***g***	**LB**	**UB**	**Mean**	**SD**	**LB**	**UB**	**Mean**	**SD**	**LB**	**UB**	**SDD%**	**>**
Right Cut (s)	2.344	0.172	2.065	0.095	<0.001**	-1.90	-3.08	-0.72	-11.7	4.3	-16.2	-7.7	-0.279	0.115	-0.375	-0.182	5.0	yes
Left Cut (s)	2.342	0.130	2.128	0.119	<0.001**	-1.63	-2.76	-0.50	-9.1	3.8	-12.5	-5.8	-0.215	0.097	-0.296	-0.133	6.4	yes
Right CODD (s)	0.466	0.121	0.214	0.068	0.001**	-2.43	-3.72	-1.14	-51.5	17.4	-72.8	-35.3	-0.252	0.129	-0.360	-0.144	25.9	yes
Left CODD (s)	0.464	0.121	0.277	0.095	_0.012*_	-1.63	-2.76	-0.50	-39.1	21.2	-54.3	-26.3	-0.187	0.123	-0.290	-0.084	33.0	yes
10-m sprint (s)	1.878	0.081	1.850	0.098	0.328	-0.29	-1.27	0.70	-1.4	3.9	-2.6	-0.4	-0.027	0.075	-0.034	0.090	5.0	no
CMAS Right	6.3	1.8	4.8	1.4	0.025*	-0.85	-1.88	0.17	-22.5	15.7	-31.5	-15.2	-1.5	1.1	-2.6	-0.4	21.7	yes
CMAS Left	6.3	1.6	4.1	1.2	0.018*	-1.46	-2.57	-0.36	-33.9	20.3	-47.3	-22.9	-2.2	1.5	-3.5	-1.0	29.3	yes

Key: IG: Intervention group; CODD: Change of direction deficit; CMAS: Cutting movement assessment score; LB: Lower bound 95% confidence interval; UB: Upper bound 95% confidence interval; SDD: Smallest detectable difference; *: $p \leq 0.05$; **: $p \leq 0.001$. Note: Italic denotes non-parametric

Table 7.5. CG pre-to-post changes

Variable	CG Pre		CG Post		Hedges' *g* effect size				% change				Mean difference				Change greater than pre SDD	
	Mean	**SD**	**Mean**	**SD**	***p***	***g***	**LB**	**UB**	**Mean**	**SD**	**LB**	**UB**	**Mean**	**SD**	**LB**	**UB**	**SDD%**	**>**
Right Cut (s)	2.395	0.131	2.251	0.089	<0.001**	-1.21	-2.28	-0.15	-5.9	3.5	-8.3	-3.7	-0.144	0.089	-0.203	-0.084	5.9	yes
Left Cut (s)	2.429	0.087	2.306	0.097	0.002*	-1.27	-2.34	-0.19	-5.0	3.9	-7.1	-3.1	-0.123	0.099	-0.190	-0.057	6.9	no
Right CODD (s)	0.510	0.117	0.365	0.088	<0.001**	-1.32	-2.41	-0.24	-27.9	10.5	-38.3	-18.5	-0.145	0.057	-0.183	-0.106	28.3	no
Left CODD (s)	0.545	0.082	0.420	0.116	0.013*	-1.17	-2.23	-0.11	-21.7	21.4	-30.8	-14.9	-0.124	0.136	-0.216	-0.032	31.5	no
10-m sprint (s)	1.885	0.058	1.885	0.075	0.957	0.01	-0.97	0.99	0.1	3.3	-0.9	1.0	0.001	0.061	-0.041	0.040	5.1	no
CMAS Right	5.1	1.5	5.2	1.5	0.779	0.08	-0.90	1.06	5.6	28.2	-11.1	22.3	0.1	1.4	-0.9	1.1	28.3	no
CMAS Left	4.6	1.4	5.0	1.1	0.306	0.33	-0.65	1.32	18.5	39.8	-5.0	42.0	0.5	1.3	-0.5	1.4	33.9	no

Key: CG: Control group; CODD: Change of direction deficit; CMAS: Cutting movement assessment score; LB: Lower bound 95% confidence interval; UB: Upper bound 95% confidence interval; SDD: Smallest detectable difference; *: $p \leq 0.05$; **: $p \leq 0.001$. Note: Italic denotes non-parametric

7.4.3.1 Right cut

Large and significant main effects of time were found for right cut completion time ($p < 0.001$, $\eta^2 = 0.829$, power = 1.000). In addition, a large and significant interaction effect of time and group for right cut completion time was also observed ($p = 0.010$, $\eta^2 = 0.330$, power = 0.779), with the IG showing significantly faster post-intervention completion times ($p < 0.001$, $g = -1.94$) compared to the CG. Moreover, large and significant improvements in right cut completion times were observed for the IG ($p < 0.001$, $g = -1.90$, -11.7%, -0.279 s) and CG ($p < 0.001$, $g = -1.21$, -5.9%, -0.144 s) (Tables 7.4 & 7.5, Figure 7.3) post-intervention, which were greater than the pre and post SDD. Mean ($p = 0.010$, $g = -1.29$) and percentage ($p = 0.005$, $g = -1.41$) improvements were significantly greater for the IG compared to CG, with large effect sizes (Figure 7.3).

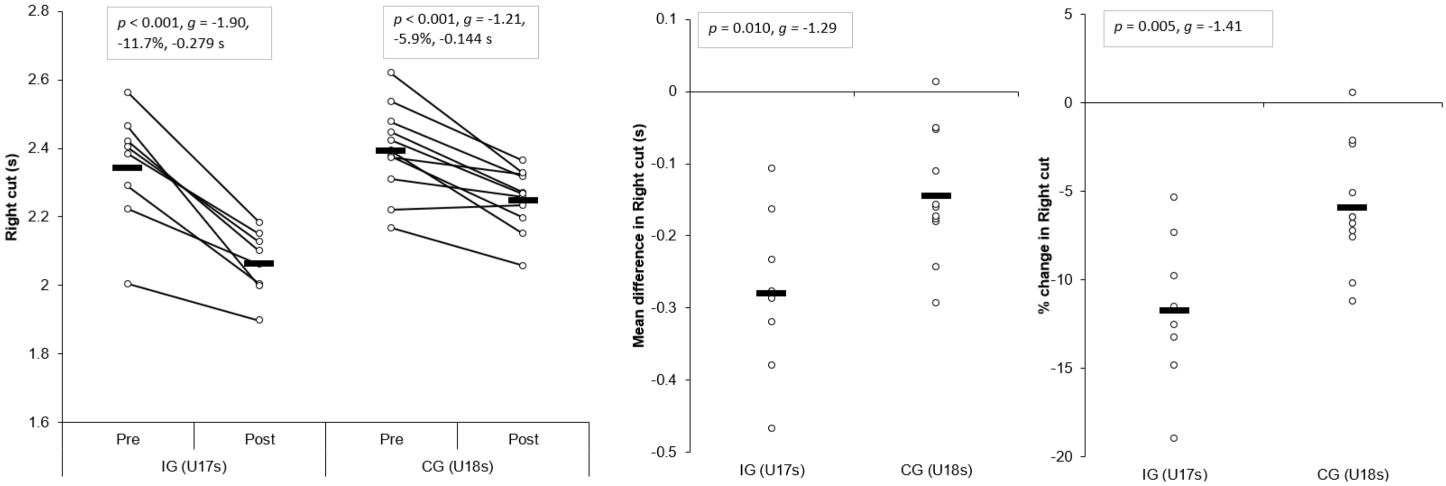


Figure 7.3. Individual plots illustrating pre-to-post changes in right cut completion times with individual mean and percentage changes; IG: Intervention group; CG: Control group. Black rectangle denotes mean.

7.4.3.2 Left cut

Large and significant main effects of time were found for left cut completion time ($p < 0.001$, $\eta^2 = 0.763$, power = 1.000). In addition, a large, yet non-significant interaction effect of time and group for left cut completion time was also observed ($p = 0.062$, $\eta^2 = 0.190$, power = 0.470); however, the IG demonstrated significantly faster post-intervention left cut completion times ($p = 0.002$, $g = -1.57$) compared to the CG. Moreover, large and significant improvements in left cut completion times were observed for the IG ($p < 0.001$, $g = -1.63$, -9.1%, -0.215 s) and CG ($p = 0.002$, $g = -1.27$, -5.0%, -0.123 s) (Tables 7.4 & 7.5, Figure 7.4) post-intervention, but these improvements were greater than pre and post SDD for the IG only. Mean ($p = 0.062$, $g = -0.89$) and percentage ($p = 0.038$, $g = -1.00$) improvements were greater for the IG compared to CG, with moderate effect sizes (Figure 7.4).

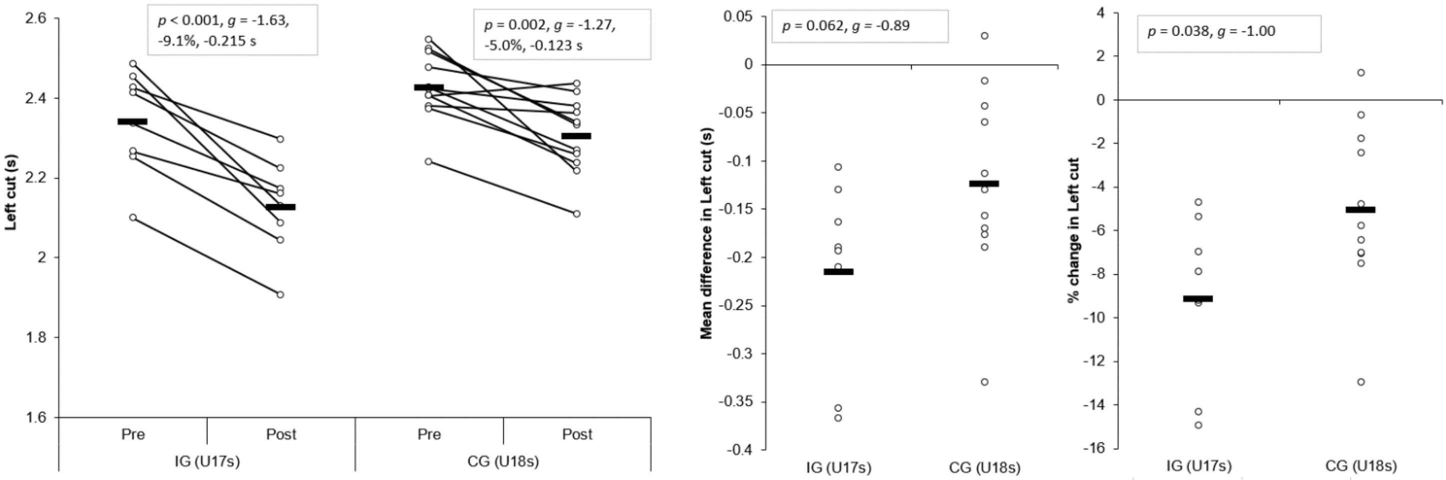


Figure 7.4. Individual plots illustrating pre-to-post changes in left cut completion times with individual mean and percentage changes; IG: Intervention group; CG: Control group. Black rectangle denotes mean.

7.4.3.3 Right CODD

Large and significant main effects of time were found for right CODDs ($p < 0.001$, $\eta^2 = 0.839$, power = 1.000). In addition, a large and significant interaction effect of time and group for right CODD was also observed ($p = 0.025$, $\eta^2 = 0.262$, power = 0.639), with the IG displaying significantly shorter post-intervention right CODDs ($p = 0.001$, $g = -1.79$) compared to the CG. Moreover, a very large and large, significant improvement in right CODD was observed for the IG ($p = 0.001$, $g = -2.43$, -51.5%, -0.252 s) and CG ($p < 0.001$, $g = -1.32$, -27.9%, -0.145 s) post-intervention, respectively (Tables 7.4 & 7.5, Figure 7.5). These changes were greater than the pre and post SDD for the IG only. Mean ($p = 0.025$, $g = -1.09$) and percentage ($p = 0.003$, $g = -1.57$) improvements were significantly greater for the IG compared to the CG, with moderate and large effect sizes (Figure 7.5), respectively.

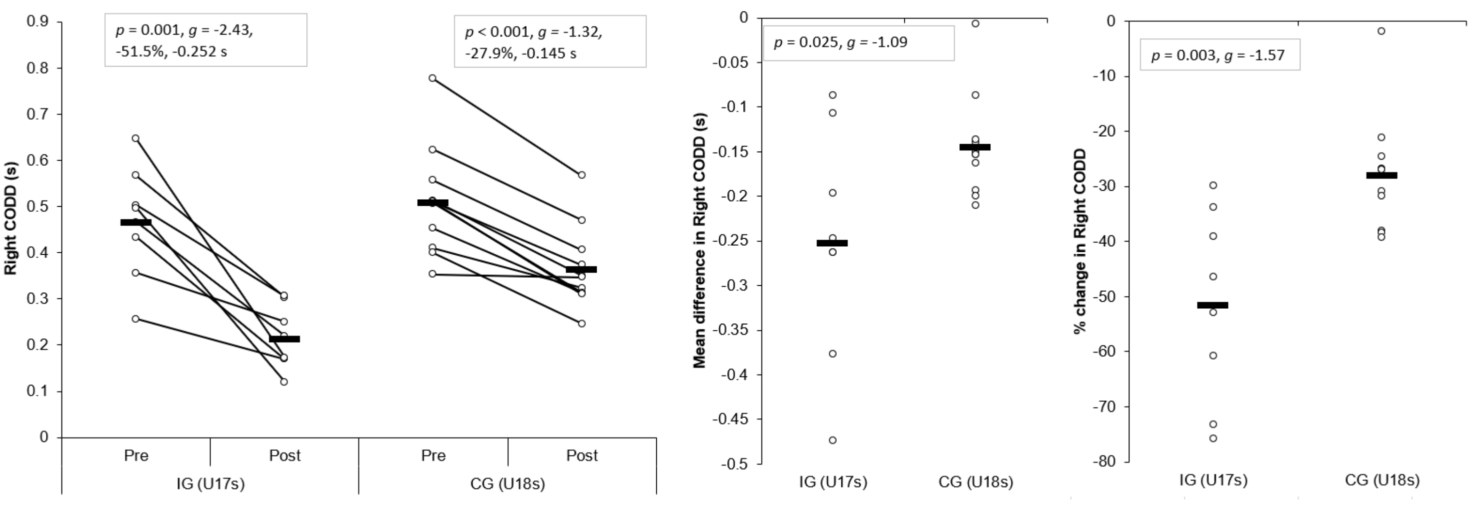


Figure 7.5. Individual plots illustrating pre-to-post changes in right CODDs with individual mean and percentage changes; IG: Intervention group; CG: Control group; CODD: Change of direction deficit. Black rectangle denotes mean.

7.4.3.4 Left CODD

Large and significant main effects of time were found for left CODDs ($p < 0.001$, $\eta^2 = 0.606$, power = 0.998). Although a small, non-significant interaction effect of time and group for left CODD was observed ($p = 0.316$, $\eta^2 = 0.059$, power = 0.164), the IG displayed significantly shorter post-intervention left CODDs ($p = 0.011$, $g = -1.27$) compared to the CG. Moreover, large and moderate, significant improvements in left CODDs was observed for the IG ($p = 0.012$, $g = -1.63$, -39.1%, -0.187 s) and CG ($p = 0.013$, $g = -1.17$, -21.7%, -0.124 s) post-intervention, respectively (Tables 7.4 & 7.5, Figure 7.6). These changes were greater than the pre and post SDD for the IG only. Although non-significant, mean ($p = 0.316$, $g = -0.46$) and percentage ($p = 0.062$, $g = -0.78$) improvements were slightly greater for the IG compared to the CG, with a small and moderate effect size (Figure 7.6), respectively.

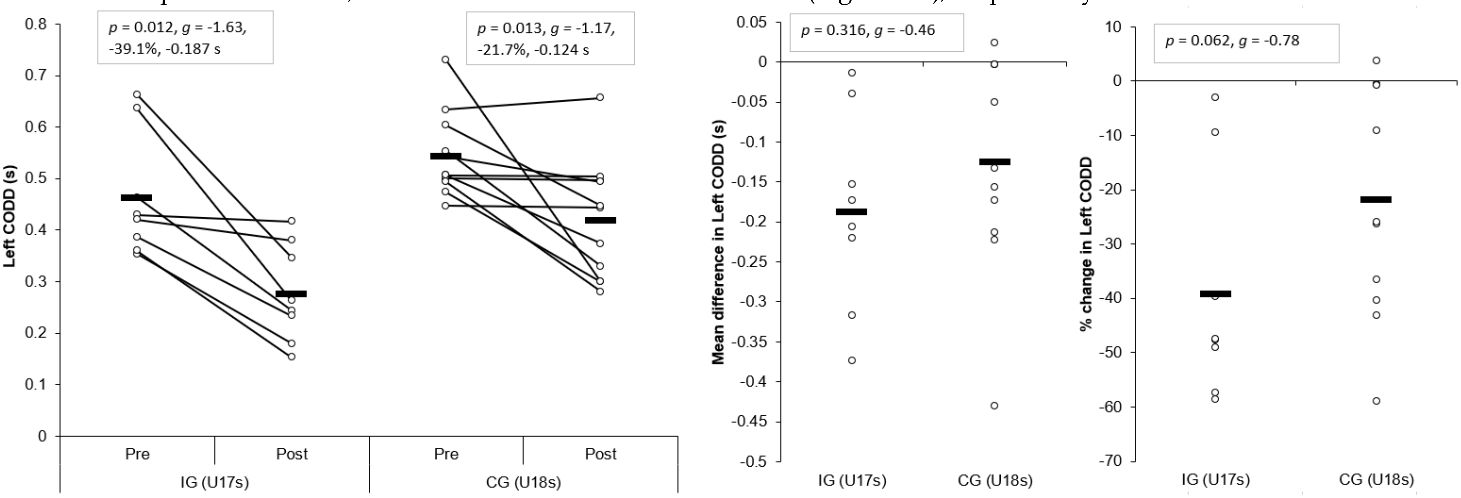


Figure 7.6. Individual plots illustrating pre-to-post changes in left CODDs with individual mean and percentage changes; IG: Intervention group; CG: Control group; CODD: Change of direction deficit. Black rectangle denotes mean.

7.4.3.5 10-m sprint

No significant main effects ($p = 0.400$, $\eta^2 = 0.042$, power = 0.129) or interactions ($p = 0.367$, $\eta^2 = 0.048$, power = 0.141) were observed for 10-m sprint performance, and no significant differences in post-intervention 10-m sprints times were demonstrated between groups ($p = 0.390$, $g = -0.39$). The IG displayed a non-significant, yet small improvement in 10-m sprints following the intervention ($p = 0.328$, $g = -0.29$, -1.4%, -0.027 s), while a non-significant, trivial difference was observed for the CG ($p = 0.957$, $g = 0.01$, +0.1%, +0.001 s) (Tables 7.4 & 7.5). These changes, however, were not greater than the SDD for both groups. Although non-significant, mean ($p = 0.374$, $g = -0.41$) and percentage ($p = 0.370$, g = -0.41) improvements were slightly greater for the IG compared to CG, with a small effect size.

7.4.3.6 CMAS right

Large and significant main effects of time were found for right CMAS (p = 0.041, η^2 = 0.224, power = 0.551). In addition, a large, significant interaction effect of time and group for right CMAS was also observed (p = 0.018, η^2 = 0.287, power = 0.694), with the IG showing slightly lower post-intervention right CMASs (p = 0.001, g = -0.26) compared to the CG. A moderate and significant improvement in right CMAS was observed for the IG (p = 0.025, g = -0.85, -22.5%, -1.46 score) following the intervention, whereas the CG demonstrated no significant and trivial changes in right CMAS post-intervention (p = 0.779, g = 0.08, +5.6%, +0.12 score) (Tables 7.4 & 7.5, Figure 7.7). These changes were greater than the pre SDD for the IG only. Although non-significant, mean (p = 0.122, g = -1.16) and percentage changes (p = 0.089, g = -1.12) were greater for the IG compared to CG, with moderate effect sizes (Figure 7.7).

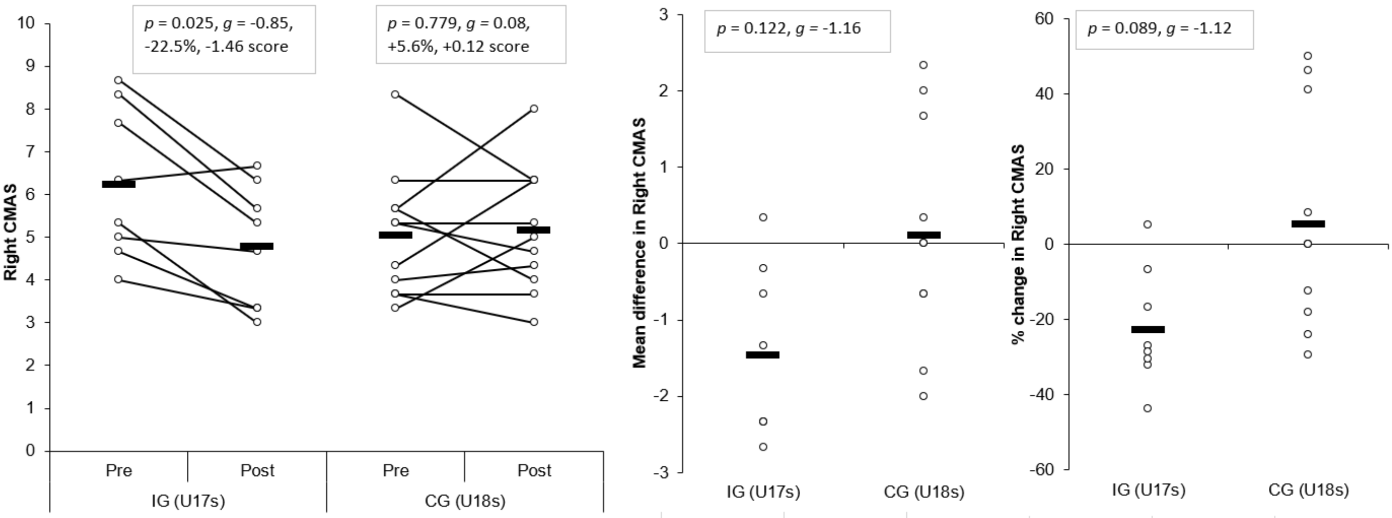


Figure 7.7. Individual plots illustrating pre-to-post changes in right CMASs with the individual mean and percentage changes; IG: Intervention group; CG: Control group; CMAS: Cutting movement assessment score. Black rectangle denotes mean.

7.4.3.7 CMAS left

Large and significant main effects of time were found for left CMAS (p = 0.015, η^2 = 0.302, power = 0.725). In addition, a large and significant interaction effect of time and group for left CMAS was also observed (p = 0.001, η^2 = 0.499, power = 0.972), with the IG showing lower post-intervention left CMASs (p = 0.318, g = -0.77) compared to the CG. A large, significant improvement in left CMAS was observed for the IG (p = 0.018, g = -1.46, -33.9%, -2.21 score) following the intervention, whereas the CG demonstrated no significant and small changes in left CMAS post-intervention (p = 0.306, g = 0.33, +18.5%, +0.45 score) (Tables 7.4 & 7.5, Figure 7.8). These changes were greater than the pre SDD for the IG only. Although non-significant, mean (p = 0.089, g = -1.83) and percentage changes (p = 0.137, g = -1.51) were substantially greater for the IG compared to CG, with large effect sizes (Figure 7.8).

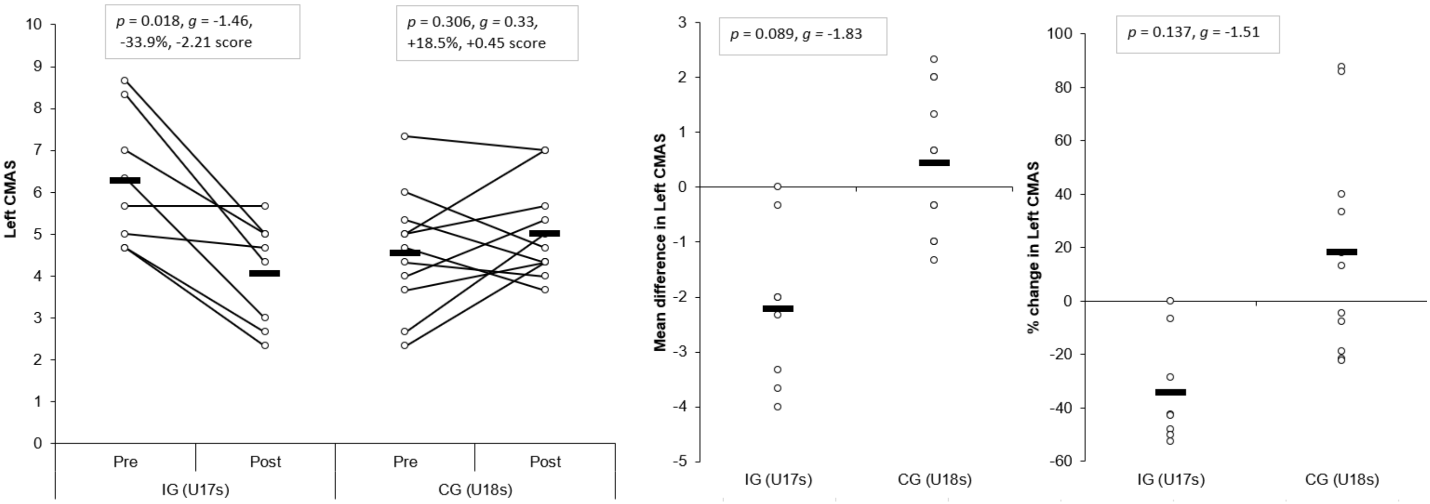


Figure 7.8. Individual plots illustrating pre-to-post changes in left CMASs with the individual mean and percentage changes; IG: Intervention group; CG: Control group; CMAS: Cutting movement assessment score. Black rectangle denotes mean.

7.5.4 Task-specific changes in cutting movement quality: CMAS deficit changes

Task-specific pre-to-post changes in cutting movement quality for the IG and CG are presented in Table 7.6 and Figure 7.9. In general, the IG demonstrated lower incidences/frequencies of CMAS deficits post-intervention, with improvements in cutting movement quality and reductions in potentially hazardous deficits including lateral trunk flexion, extended knee postures, knee valgus, hip internal rotation, and improved PFC braking strategies (Table 7.7, Figure 7.9). Conversely, changes in the frequencies of CMAS deficits were smaller for the CG, and in some cases frequencies of CMAS deficits increased, including lateral trunk flexion, extended knee postures, trunk leaning back, and foot position (Table 7.7, Figure 7.9).

Table 7.6. Task-specific pre-post changes in CMAS deficits

CMAS criteria	IG – Pre-to-post changes in CMAS criteria				CG – Pre-to-post changes in CMAS criteria			
	Right cut		Left cut		Right cut		Left cut	
	n of deficits	%	n of deficits	%	n of deficits	%	n of deficits	%
No clear PFC braking strategy	-5	-24	-12	-57	-8	-33	-2	-8
Wide foot plant	3	33	-2	-22	1	8	2	22
Hip internal rotation	-7	-100	-5	-100	-1	-25	-1	-33
Initial knee valgus	-9	-39	-5	-28	-2	-9	-2	-11
Internal/external foot	3	27	-5	-26	1	6	9	60
Upright trunk	2	22	3	50	-5	-42	1	9
Lateral trunk flexion	-4	-27	-5	-28	8	67	-1	-5
Trunk leaning back	-4	-57	-5	-63	4	133	5	250
Limited knee flexion	-1	-17	-5	-50	-2	-13	6	50
Excessive knee valgus	-12	-67	-5	-50	-3	-19	-3	-43

Key: CMAS: Cutting movement assessment score; PFC: Penultimate foot contact; IG Intervention group; CG: Control group; n: number

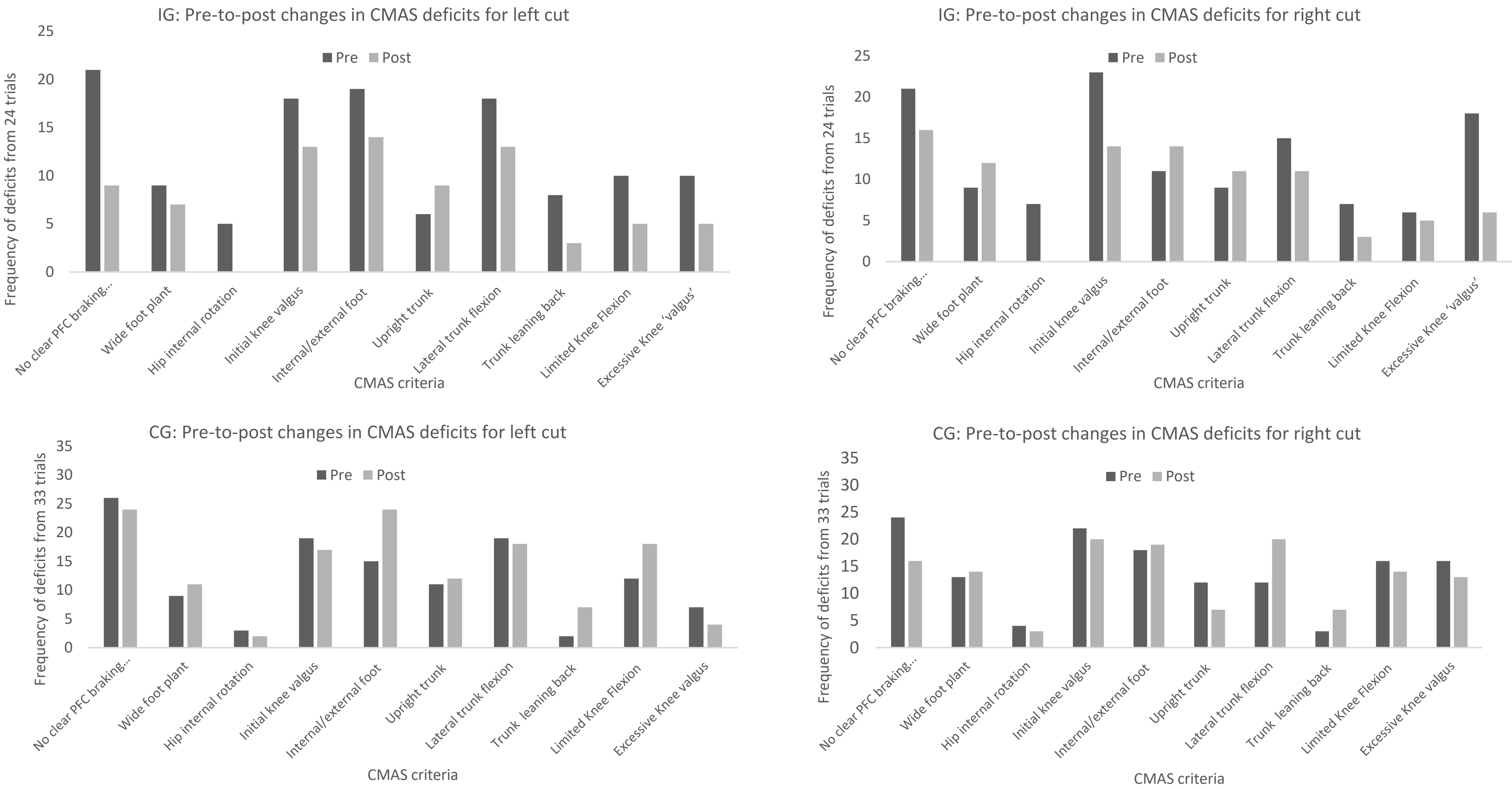


Figure 7.9. Cutting task-specific changes in CMAS deficits

7.5 DISCUSSION

The aim of the present study was to determine the effects of a six-week COD speed and technique modification training intervention on cutting performance and movement quality in male youth soccer players. The primary findings were that six-weeks COD speed and technique modification training performed in-season, in addition to normal skills and strength training, produced meaningful improvements in cutting performance times, CODDs, and cutting movement quality (i.e., lower CMAS) in male youth soccer players (Tables 7.4 & 7.5, Figures 7.3-7.8), thus supporting the study hypotheses. The observed improvements in performance time and CODDs in the IG were on average two times greater than the CG, who also continued their normal warm-ups (Tables 7.4 & 7.5). However, the CG demonstrated no meaningful or significant improvements in CMASs, in contrast to the IG who demonstrated meaningful improvements in cutting movement quality.

As cutting actions are frequently performed manoeuvres in soccer (51, 487) and linked to decisive moments (i.e., assists and goal scoring) (171), the ability to cut rapidly can be considered an important quality to develop. The results of the present study substantiate the findings of previous research that found COD speed training interventions improved COD speed completion times in male youth soccer players (77, 81, 87, 347, 348) and college athletes (332, 630), with the present study finding COD speed and technique modification training resulted in meaningful improvements in cutting completion time ($p < 0.001$, $g = 1.63$-1.90, ~9-11%), and these changes were greater than the CG (Tables 7.4 & 7.5, Figures 7.3 & 7.4) and pre and post SDD. Moreover, CODD has been recently developed and suggested to provide a more isolated measure of COD ability (113, 411-413). Previous studies which have shown improvements in COD speed tasks have only assessed completion times (77, 81, 87, 347, 348), and these tasks are mainly comprised of linear running and thus, biased towards athletes with superior acceleration and linear speed capabilities (329, 331, 412, 413, 500, 576). Conversely, to the best of the author's knowledge, the present study is the first to monitor changes in CODD in response to a training intervention in male youth soccer players. Critically, substantial and meaningful improvements in CODD were demonstrated by the IG ($p \leq 0.012$, $g = -1.63$ to- 2.43, ~40-52%) which were approximately two times greater than the CG (Tables 7.4 & 7.5, Figures 7.5 & 7.6). These findings highlight the effectiveness of six-weeks field-based COD speed and technique modification training in male youth soccer athletes, which can be achieved in-season with two, twenty-minute sessions per week. This form of training can be simply and easily integrated into the warm-ups of field-based tactical/technical sessions in soccer, highlighting the applicability and feasibility of COD speed and technique modification training.

Cutting is a key action associated with non-contact ACL injuries in soccer (63, 204, 583) due to the propensity to generate large multiplanar knee joint loads (39, 126, 127, 273, 301) that can strain (39, 126, 127, 273, 301) and potentially rupture the ACL (327, 365). COD technique modification training has been shown to be an effective modality for addressing "high-risk" postures (lateral trunk flexion, lateral foot plant distance, IFPAs) and reducing potentially hazardous knee joint loading (76, 126, 277). Although the present study used a qualitative screening tool (CMAS) to monitor changes in cutting movement quality, the CMAS has been recently validated against the gold standard of 3D motion analysis with strong relationships observed between CMAS and peak KAM (278) (Chapter 6), which presents as a more practical, less expensive screening tool to implement in applied sporting environments (187, 476, 478, 480). In addition, the effectiveness of neuromuscular training interventions on jump-landing mechanics have been evaluated using qualitative screening tools such as the LESS (135-137, 423, 438) and TJA (295, 540). Nevertheless, the results of this study confirm that COD speed and technique modification training (with feedback and externally directed verbal cues from a coach) resulted in meaningfully lower CMASs post-intervention for the IG which were greater than the pre SDD (Tables 7.4 & 7.5, Figures 7.7 & 7.8) ($p \leq 0.025$, g = -0.85 to -1.46, -23 to -34% vs. +6-19%), while the CG remained unchanged.

Indeed, the CG did demonstrate improvements in cutting completion times and CODDs, albeit not to the same extent as the IG. However, it is important to note that the CG did not abstain from training and still performed sprint and low-level plyometric training (to maintain ecological validity) while the IG performed the COD speed and technique modification training. Both groups performed similar strength training programmes and, as they were adolescents, were still biologically maturing. With this in mind, it is unsurprising that the CG did improve cutting performance, but critically did not improve cutting movement quality in contrast to the IG; highlighting the effectiveness of COD speed and technique modification training.

Similar to Stroube et al. (540), which assessed task-specific changes in the TJA (i.e., changes in the frequencies of deficits demonstrated) following a neuromuscular training intervention with task-specific feedback, the IG in the present study demonstrated improved cutting movement quality and reductions in frequencies in "high-risk" deficits including lateral trunk flexion, extended knee postures, knee valgus, and improved PFC braking strategies (Tables 7.5-7.7, Figures 7.7-7.9). These findings are noteworthy because these aforementioned "high-risk" postures are associated with increased knee joint loading (127, 189, 228, 267, 272, 273, 301, 358, 516, 519) which increases ACL (242, 298, 342, 343, 512, 614) and PFP injury risk (393, 395, 580). Furthermore, these postures are also characteristics of non-contact ACL injury (63, 94, 204, 248, 270, 291, 298, 310, 426, 541, 583). As such, six-weeks COD speed training and technique modification with externally focused coaching cues and feedback from a coach is an effective training modality for reducing "high-risk" biomechanical and neuromuscular control deficits, and overall improving movement quality in male youth soccer players. It is worth noting, however, that CMAS deficits are still demonstrated by male youth soccer players (Table 7.7, Figure 7.9) and thus, they should continue performing injury mitigation training interventions to address these deficits (241, 432, 435, 479).

It has been suggested a "performance-injury conflict" could exist when modifying COD technique (148, 183, 228) (Chapter 5), whereby addressing "high-risk" postures could be detrimental to performance (115, 148, 183, 228). While Dempsey et al. (126) reported reductions in knee joint loads due to changes in lateral foot plant distance and trunk position, the authors failed to consider the implications of such changes on performance (i.e., completion time, GCT, and exit velocity). Jones et al. (277) found changes in pivoting technique resulted in lower KAMs and faster completion times in netball players. However, these studies have used 3D motion analysis to monitor changes in COD technique and more importantly, these studies have not contained a CG; therefore, the results should be interpreted with caution. To the best of the author's knowledge, the present study is the first to consider the effect of COD speed and technique modification training on both performance (completion times, CODD) and injury risk (CMAS) in comparison to a CG, using a field-based screening tool which was performed in a "real-world" setting. Notably, COD speed and technique modification training with feedback and external verbal coaching cues was effective in improving cutting performance and movement quality (Tables 7.4 & 7.5), and these were significantly and meaningfully greater than the CG and SDD (Figures 7.3-7.8). Collectively, these findings highlight that COD speed and technique modification in male youth soccer players is an effective training modality for enhancing performance and addressing movement deficits associated with increased knee joint loading and potential injury risk. This finding is noteworthy because, since the introduction of the elite performance player plan, injury rates in youth-soccer have increased three-fold (481) and as such, reducing biomechanical and neuromuscular risk factors in youth soccer is considered highly important, particularly as these players are striving for professional contracts.

The COD speed and technique modification training intervention focused on modifying biomechanical deficits associated with increased knee joint loads (146, 148, 183) and promoting techniques required for faster performance (146, 183, 228, 346). For example, the programme focused on several aspects: a wide foot-plant is required for MLPF generation and subsequent exit velocity during cutting (228, 261, 272); faster performance and lower knee joint loading has been associated with increased PFC braking forces (143, 147, 202, 273, 275); and trunk lean towards the direction of travel and reduced lateral trunk flexion is associated with faster performance (183, 229, 346) and reduced knee joint loads (127, 141, 189, 267, 272). Moreover, knee valgus is also a hazardous "high-risk" posture (272, 274, 298, 301, 358, 516) with no associated performance benefits (183, 228, 346) and thus, was a further desired technical change in response to the intervention. Consequently, the IG training programme consisted of COD drills and externally focused verbal coaching cues ("slam on the brakes early", "push the ground away" and "lean/face towards the direction of travel") to promote faster performance (35, 461, 610), safer mechanics (34, 598), and to improve motor skill retention (88, 169, 598, 620). These cues were used to evoke technical changes to encourage PFC braking and trunk lean towards the direction of travel which are techniques associated with faster performance and reduced knee joint loads, while MLPF was also emphasised to promote faster exit velocities. Interestingly, post-intervention, the IG demonstrated lower CMASs (Table 7.5, Figures 7.7 & 7.8), which can be attributed to reduced incidences of CMAS deficits, such as lateral trunk flexion, initial and excessive knee valgus, hip internal rotation, and lack of PFC braking (Table 7.7, Figure 7.9), while a wide lateral foot plant remained relatively unchanged. Although these were qualitative evaluations only, it is speculated that the IG demonstrated safer cutting mechanics, and due to the strong relationship observed between CMAS and peak KAM (278) (Chapter 6), theoretically, the IG may demonstrate lower knee joint loading, which subsequently reduces non-contact ACL (242, 298, 342, 343, 512, 614) and PFP injury risk (393, 395, 580). Further research is required to determine the effect of COD technique modification on knee joint loading and performance using 3D motion analysis to further substantiate this claim.

7.6 LIMITATIONS

It should be noted that the present study only investigated a 70˚ side-step cutting task from a short approach distance (5-m), and thus, is only reflective of low-entry velocity side-step cutting ability. As the biomechanical demands of COD are angle- and velocity-dependent (148, 411), and other COD actions are also performed, such as XOCs, split-steps and pivots, further research is needed to determine the effects of COD speed and technique modification training on CODs from different angles and approach distance, while also investigating different types of COD actions. Additionally, it is worth noting that elite male youth soccer players were only investigated who are potentially not as technically competent at COD and have lower training ages and physical conditioning compared to senior, experienced athletes, and therefore potentially possess greater scope for improvements in terms of cutting performance and movement quality. As such, caution is advised regarding the generalisation of these results to senior soccer players and different athletic populations. Further insight is required into the effects of COD speed and technique modification training in senior soccer players and different athletic populations where cutting is a highly prevalent action for performance and also non-contact ACL injury, such as rugby (376, 601, 632), American football (270), and handball (426, 552).

It should be noted that there were five dropouts for the IG (Figure 7.1) due to match-related injuries or illness. This is similar to previous COD speed research that also documented dropouts in their training interventions (348, 630). However, the present study still achieved *a priori* sample size statistical power recommendation and dropouts reflect the environment of performing an in-season training intervention in a "real-world" professional soccer setting. Finally, while improved cutting movement quality was demonstrated following the COD speed and technique modification training, the short- and long-term retention benefits of this training are unknown, and therefore a recommended area of future research.

7.7 CONCLUSION

In conclusion, six-weeks COD speed and technique modification training performed in-season (with the use of external verbal coaching cues and feedback), in addition to normal skills and strength training, resulted in significant and meaningful improvements in cutting completion time and CODDs in male youth soccer players. These changes in performance were approximately two times greater than the CG and exceeded the SDD. Furthermore, COD speed and technique modification training resulted in lower CMASs and improved cutting movement quality, while the CG demonstrated no meaningful or significant changes in CMAS. The improvements in cutting movement quality observed for the IG were attributed to technical improvements (reductions in deficits), such as lateral trunk flexion, knee valgus, PFC braking, and internal hip rotation and overall cutting movement quality. These findings indicate that improvements in cutting performance and movement quality can be achieved in-season, in a "real-world" sporting environment. As such, practitioners working with male youth soccer players should consider implementing two twenty-minute COD speed and technique modification training sessions a week, in addition to normal skills and strength training, for improvements in cutting movement quality and performance.

CHAPTER 8: Biomechanical Effects of a Six-Week Change of Direction Technique Modification Intervention: Implications for Performance and Injury Risk

8.1 INTRODUCTION

Change of direction ability is a fundamental movement associated with successful performance in multidirectional sports (51, 171, 283, 409, 487, 510, 549, 628), such as evading an opponent (402, 601, 632) or moving into space to receive a pass (184). Changing direction, however, has also been identified as a key action associated with non-contact ACL injuries in numerous multidirectional sports (soccer, rugby, handball, netball, Australian rules football, American football, and badminton) (52, 63, 94, 170, 270, 291, 298, 376, 426, 583) due to the propensity to generate high multiplanar knee joint loading (flexion, rotation, and abduction loading) during the plant foot contact (39, 126, 127, 273, 301), thus increasing ACL strain (26, 288, 342, 424, 513). ACL injuries are a debilitating injury with short- and long-term consequences (financial, health, and psychological) (112, 241, 315, 333, 468), with an elevated and earlier risk of developing osteoarthritis a primary concern (333, 571). Annual ACL injury rates are estimated to be 250,000 in the United States (241) with two million injuries worldwide (483), and over one billion dollars, estimated, spent annually on the reconstruction and rehabilitation in the United States. ACL injuries typically require surgery (210); thus, extensive rehabilitation periods are required, resulting in prolonged absence and the potential to lose sporting scholarships or contracts (375), while athletes who do successfully return to sport post ACL reconstruction, may demonstrate reduced sports-related performance, reduced number of appearances, and shorter career longevity (313, 368). Therefore, understanding the mechanics, techniques, and training interventions that can mitigate injury risk during COD actions, while improving performance, are of great interest to practitioners working with multidirectional athletes.

Although ACL injury risk factors are multifactorial (anatomical, hormonal, biomechanical, neuromuscular, environmental) (241, 469), ACL injuries occur when a load is applied that exceeds the ligaments' tolerance (327, 365); thus, in order to reduce ACL injury risk, particularly non-contact ACL injuries, an effective strategy is to modify an athlete's movement mechanics (addressing biomechanical and neuromuscular deficits) to reduce the magnitude of knee joint loading through biomechanically and neuromuscular informed training interventions (126, 183, 241, 245, 246, 391, 435, 508, 535). COD techniques with a wide lateral foot plant (127, 228, 272, 301), greater hip abduction angles (519, 593), increased internal IFPAs (274, 519), increased initial hip internal rotation angles (228, 358, 516, 519), greater peak and initial KAAs (272, 274, 301, 358, 516), reduced knee flexion (76, 593), greater lateral trunk flexion (127, 189, 267, 272), and greater GRFs (273, 516, 519) are associated with greater KAMs and thus injury risk (242, 342, 343, 512, 614). Additionally, wide lateral foot plant distances, trunk rotation towards the stance limb, trunk flexion displacements, and hip internal rotation moments are associated with greater KIRMs (127, 189), which when combined with KAMs produces greater ACL strain (multiplanar) compared to uniplanar loading (26, 288, 342, 424, 513). As such, addressing and modifying the aforementioned variables associated with KAMs and KIRMs could be an effective strategy for reducing ACL loading and thus potential ACL injury risk during COD (183).

There is evidence, although limited, which indicates the techniques and mechanics required for faster COD performance are in direct conflict with techniques and mechanics required for safer (i.e., lower knee joint loads) directional changes (115, 183, 228) (Chapter 5). For instance, COD techniques, such as increased IFPAs and pelvic and hip internal rotation angles are associated with greater KAMs (127, 274, 519), but may be optimal for COD performance due to effective alignment of the whole-body COM into the new intended direction (229, 274). Lateral trunk flexion has been shown to increase knee joint loading (126, 127, 272); however, this strategy may be adopted by athletes to deceive (feint) opponents (61, 62, 252). Faster COD performance has been associated with greater KFMs (228) and posterior GRF (143, 202, 275), but these two variables are also associated with proximal anterior tibial shear (506, 631) and potential ACL loading (42, 43, 343, 615). Importantly, wide lateral foot plant distances (126, 127, 228, 272, 301) are also associated with greater KAMs, where larger moment arms and KAMs are created with a more medial whole-body position with respect to the foot and COP positioning more lateral to the COM of the body and tibia (228, 274). However, a wide lateral foot plant is required to generate MLGRF and impulse to accelerate into the new direction (228, 261, 272, 597), and is a key determinant of faster cutting performance.

Havens and Sigward (228) revealed faster cutting performance was associated with greater lateral foot plant distances, ML impulse, and internal hip rotation angle, though it is worth noting that greater KAMs were also observed with wider lateral foot plants, which may increase ACL loading. Results from Chapter 5 substantiate Havens and Sigward (228), by confirming that techniques and mechanics associated with faster performance (i.e., faster velocities over key instances of the PFC and FFC, greater FFC braking forces over short GCTs, greater FFC KFMs, smaller hip and KFAs and ROM, wider lateral foot plants, and greater internal IFPAs) are at odds with reduced knee joint loading. Collectively, these studies suggest that there is a “performance-injury conflict” during COD, which is problematic for practitioners who seek to improve their athletes’ performance and reduce risk of injury. Further insight is required to confirm whether simultaneous reductions in knee joint loads and improvements in cutting performance can be achieved.

Based on the scoping review in Chapter 2.3.5, COD technique modification training is an effective training strategy for reducing “high-risk” COD mechanics and subsequent knee joint loads (76, 115, 126, 277). Reducing knee joint loads can be achieved via reducing the magnitude of the moment arm (particularly in the frontal plane), GRF, or a combination of the two (301). Decreases in frontal and transverse knee joint loads during cutting have been demonstrated as a result of acute (127) and chronic (126) COD technique modification via alterations in lateral foot plant distance and orientation, and trunk alignment; however, these studies have failed to acknowledge measurement error when interpreting findings, and the chronic six-week intervention study did not have a CG. Thus, it is uncertain whether such changes were "real", and the results should therefore be treated with caution. Furthermore, both COD technique modification studies failed to consider the implications of technique modifications on performance (i.e., completion time, GCT, and exit velocity). Bringing the plant foot closer to the midline can reduce KAMs by reducing the moment arm distance but could be suboptimal for MLGRF production and may result in suboptimal COD performance (228, 261, 272, 596). As athletes are driven by performance, they may be unlikely to adopt movement strategies which decrease knee injury risk if they do not result in effective performance (183, 228).

Increasing knee flexion (acutely, within-session) during side-stepping has been shown to reduce KFMs and posterior GRF but negatively affected performance by increasing GCT and reducing exit velocity (115). Instructing athletes to move their COM closer to the their base of support (acutely, within-session) and increase KFA has been shown to reduce peak KAMs during side-step cutting (76), with no detrimental changes in average speed. Unfortunately, however, the authors did not delineate how average speed was quantified. Jones et al. (277) reported a reduction in 180° turning KAMs and improved modified 505 performance in female netball players as a result of a six-week technique modification intervention which consisted of technique drills that encouraged PFC braking, backwards trunk inclination, and a neutral foot position (i.e., reducing IFPA). Interestingly, a strong association between changes in IFPA and KAMs (r^2 = 37%, p = 0.028) was observed; however, similar to Dempsey et al. (126), there was no CG and the findings were only presented in abstract/presentation format. Finally, in Chapter 7, it was shown that six-weeks COD speed and technique modification which focused on cues to encourage earlier and greater PFC braking, trunk lean towards the intended direction of travel, and rapid and forceful push-off improved cutting performance and CMASs (movement quality) in male youth soccer players. Although these results are promising, movement quality was examined qualitatively and therefore must be further evaluated using 3D motion and GRF analysis.

Collectively, a "performance-injury conflict" may exist between improving COD performance and reducing knee joint loads whereby addressing one issue maybe to the detriment to the other and vice versa. As such, exploration into the biomechanical effects of COD technique modification on both performance and knee joint loading is warranted to improve our knowledge of the potential "performance-injury conflict" but acknowledging measurement errors and having a CG. Conducting this research may assist in the development of more effective ACL injury mitigation and COD speed programmes (183). However, it should be noted that a COD technique modification intervention which reduces potential risk of injury (i.e., reducing knee joint loading), without compromising performance (i.e., maintaining completion time, GCT, and exit velocity), could still be viewed as successful. Athletes are driven by performance and are less likely to adopt techniques which do not facilitate faster and effective performance (183, 228). To improve our understanding of COD in terms of performance and risk of injury, more detailed analysis considering both aspects are warranted. Therefore, based on the above, the overall aims of this research are two-fold:

1. To evaluate the effectiveness of a 6-week COD technique modification intervention on COD performance (completion time, GCT, and exit velocity) and injury risk multiplanar knee joint loads (KAM, KIRM, KFM) during 45° (CUT45) and 90° (CUT90) side-step cutting.
2. To identify which kinetic and kinematic factors explain changes in performance and knee joint loads.

It is hypothesised that:

1. A COD technique modification programme will concurrently reduce knee joint loads and improve COD performance in multidirectional athletes.
2. Changes in technique variables such as IFPA, lateral trunk flexion, KFA at IC, and HBF ratio will explain reductions in knee joint loads.
3. Changes in GCT, velocity at key instances of the PFC and FFC, propulsive forces, knee flexion ROM, and pelvic rotation will explain improvements in performance.

If successful (i.e., reducing knee joint loading and maintaining or improving performance), practitioners could implement the COD technique modification programme as a potential ACL injury mitigation programme without compromising COD performance in field-settings.

8.2 METHODS

8.2.1 Research design

A non-randomized, controlled 6-week intervention study with a repeated measures pre-to-post design was used. Men from multidirectional sports were recruited for the IG which consisted of a 6-week COD speed and technique modification training programme (Appendix 7.1 and 7.2), consisting of two 30-minute sessions per week in addition to their normal sports and resistance training. Based on the findings from Chapter 4 (Study 1) that KAMs, KAAs, and KIRMs, can decrease between sessions, it was central that a CG was included to examine the biomechanical effects of cutting technique modification and permit greater certainty regarding training-induced biomechanical changes. Conversely, men from multidirectional sports acted as the CG and continued their normal sports and resistance training, while abstaining from specific COD training. Pre-to-post assessments of 45° and 90° cutting performance and COD biomechanics were assessed using 3D motion and GRF analysis to monitor the effectiveness of the training intervention. This study was funded by a doctoral research grant by the NSCA for participant recruitment.

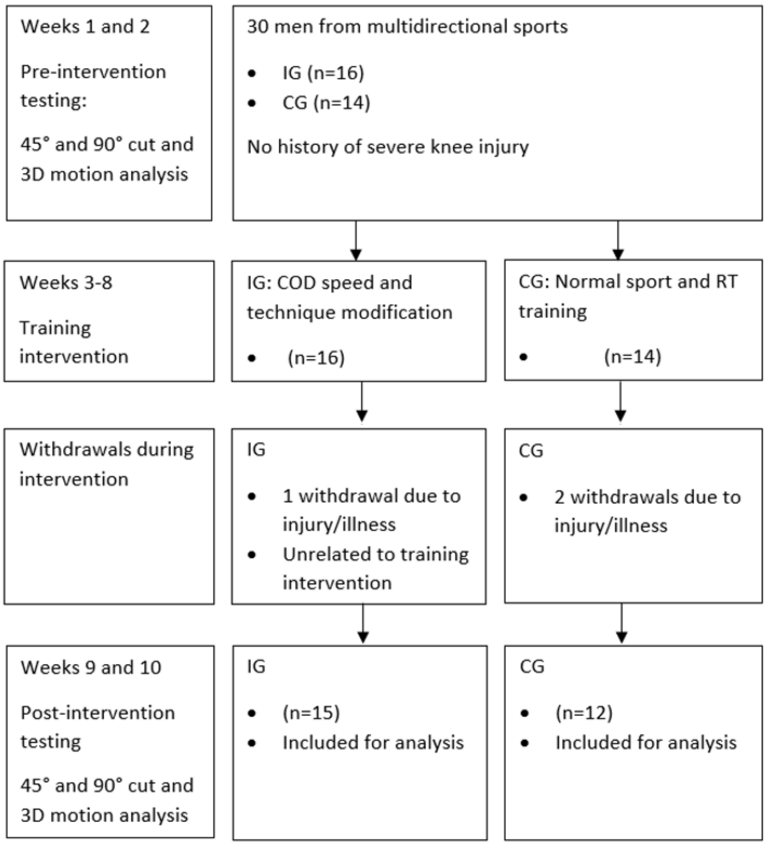


Figure 8.1. Flow diagram of participant participation throughout all stages of the intervention study. IG: Intervention group; CG: Control group; RT: Resistance training.

8.2.2 Participants

Unlike the previous intervention study (Chapter 7), it was not feasible to recruit a soccer team for the present intervention study. As such, a convenience sampling approach was adopted, whereby 30 men from multidirectional sports (21 soccer amateur / semi-professional level) were recruited who were familiar with cuts and participated in this study. Based on the work of Jones et al. (277) performed in the same laboratory for pre-to-post (dependent t-test) changes in peak KAMs during planned 180° turning performance, a minimum sample size of 14 per group was determined from an *a priori* power analysis using G*Power (Version 3.1, University of Dusseldorf, Germany) (172). This was based upon an effect size of 0.73, a power of 0.80, and type 1 error or alpha level of 0.05.

Sixteen men who participated in multidirectional sports (soccer n = 12, rugby n = 4) (age: 23.5 ± 5.2 years; height: 1.80 ± 0.05 m; mass: 81.6 ± 11.4 kg) were recruited for the IG. 14 of 16 participants for the IG stated right limb dominance (kicking and preferred push-off limb). Conversely, fourteen men who also participated in multidirectional sports (soccer n = 9, rugby n = 4, field hockey n = 1) (age: 22.2 ± 5.0 years; height: 1.76 ± 0.08 m; mass: 72.7 ± 12.4 kg) acted as the CG and continued their normal sport and resistance training sessions. All participants from the CG stated right limb dominance. The investigation was approved by the Institutional Ethics Review Board (HSR1617-131), and all participants were informed of the benefits and risks of the investigation prior to signing an institutionally approved consent form to participate in the study.

All participants from both groups possessed at least five years' training experience in their respective sport and had never sustained a knee injury such as an ACL injury prior to testing. All participants had minimum one years' resistance training experience, and all performed two resistance training sessions a week. At the time of the training intervention, all participants completed two skills sessions and played one competitive match a week. All of the procedures were carried out during the competitive season to ensure that no large physical changes were made as a result of the conditioning state (81). To remain as an active participant in the study and used for further analysis, participants were not allowed to miss more than two of the 12 sessions in total (i.e., ≥ 83% compliance rate). Subsequently, due to match-related injuries or illness, one and two participants withdrew from the IG and CG, resulting in sample sizes of 15 and 12 (Figure 8.1) respectively. Participants in the IG completed on average 11.9 ± 0.4 sessions (98.3 ± 3.5%), with 12 participants completing 12 (100%) sessions and three completing 11 sessions (91.7%).

<u>8.2.3 Procedures</u>

For a detailed description of the warm-up, short 45° and 90° cut, marker placement, and 3D motion analysis procedures, please refer to Chapter 3.2 and 3.3 in the research methodology. In brief, each participant performed six acceptable trials of the short 45° and 90° cut and were provided with standardised footwear to control for shoe-surface interface. Marker and force data were collected over the PFC and FFC, and completion time was also measured for each trial. A minimum of five trials were used in the analysis of each participant based on visual inspection of the motion files (370). Unfortunately, due to logistical and time constraints, it was not feasible to simultaneously capture 2D video footage during the 3D biomechanical testing, thus CMAS screening could not take place.

8.2.4 Cutting kinetic and kinematic variables

To assess changes in cutting performance, completion time, GCT, and exit velocity were the primary outcome variables. Approach velocity, velocity at FFC, and cut angle were also secondary variables to assess performance. To assess potential injury risk, multiplanar knee joint loads (KAM, KIRM, KFM) were evaluated over WA during the FFC for both cutting tasks and were considered the primary outcome variables and surrogates of non-contact ACL injury risk. IC and peak KAA and KRAs were considered secondary injury risk variables. GRF and joint kinetic and kinematic variables are outlined in Tables 3.1 & 3.2.

8.2.4 6-week COD speed and technique modification training intervention

A six-week COD speed and technique modification intervention described in Appendix 7.1, was performed by the IG twice a week (30 minutes per session), with minimum 48 hours between sessions. The intervention was slightly adapted from the intervention performed in Chapter 7 (Study 4) due to the ability to perform all training sessions on the same surface. Therefore, session 1 in this intervention contained more decelerations compared to the session 1 of the previous intervention (Chapter 7, Study 4), but the number of CODs were similar. The six-week technique modification intervention focused on pre-planned low intensity decelerations, cuts, and turns (weeks 1-2), before progressing intensity via velocity and angle (weeks 3-4) (56, 131, 148, 411), and introducing a stimulus with increased intensity (weeks 3-6). The COD programme was in accordance with COD speed recommendations from the NSCA (215), Nimphius (410), and recent braking strategy recommendations (147). The duration, distances, and number of CODs were similar to previously successful 6-week COD speed (77, 81, 87, 347, 348, 630) and COD technique modification studies (126, 277). The sessions were led by the principle investigator who is a certified strength and conditioning specialist with extensive experience in coaching COD speed and agility drills. All COD speed and technique modifications sessions took place in the Human Performance Laboratory, on the same surface as used for 3D motion analysis. Athlete to coach ratios were generally 5:1, but on occasions did range from 1:1 to 1:10.

The COD technique modification intervention focused on three aspects outlined in Table 9.1 based on the earlier results observed in Chapter 5 (Study 2) that confirmed that high MLPFs over short GCTs were associated with faster cutting performance, reduced lateral trunk flexion was associated with shorter GCTs, PFC dominant braking strategies were associated with faster cutting performance and lower KAMs, and greater KAAs offered no performance benefits and increased knee joint loads. Participants were given individual feedback regarding their technique, and external verbal coaching cues were used to promote safer mechanics (34, 598), promote faster performance (35, 461, 610), and facilitate better motor skill retention (88, 169, 598, 620).

Table 8.1. COD technique modification and coaching cues

Cutting technique modification	Coaching cue and rationale
Adopting a PFC dominant braking strategy	"Slam on the brakes" (to reduce GRF in cutting/turning limb (for the 90° task only)) (147)
Lower-limb frontal plane control and force expression	"Cushion and push/punch the ground away" (to enhance MLPF, reduce KAAs, and encourage active limb at touch-down) (146, 631)
Trunk control	"Face towards the direction of travel" (to reduce lateral trunk flexion) (141, 146)

Key: GRF: Ground reaction force; PFC: Penultimate foot contact; KAA: Knee abduction angle; MLPF: Medio-lateral propulsive force

8.3 STATISTICAL ANALYSES

All statistical analyses were performed using SPSS v 25 (SPSS Inc., Chicago, IL, USA) and Microsoft Excel (version 2016, Microsoft Corp., Redmond, WA, USA).

<u>8.4.1 Pre-to-post changes in primary outcome variables</u>

Normality was inspected for all variables using a Shapiro-Wilks test. A two-way mixed ANOVA (group; time) with group as a between-subjects factor measured at 2 levels (IG and CG), and time (pre- and post-training measures) the within-subject factor. This was used to identify any significant main (time) or interaction (group × time) effects for outcome variables between IG and CG, pre-to-post testing. A Bonferroni-corrected pairwise comparison design was used to further analyse the effect of the group when a significant interaction effect was observed for time and group. Partial eta squared effect sizes were calculated for all ANOVAs with the values of 0.010-0.059, 0.060-0.149, and ≥ 0.150 considered as small, medium, and large, respectively, according to Cohen (96).

Pre-to-post changes in variables for each group were assessed using paired sample t-tests for parametric data and Wilcoxon-sign ranked tests for non-parametric data. Magnitudes of differences were assessed using Hedges' *g* effect sizes with 95% CIs, and mean change. Hedges' *g* effect sizes were calculated as described previously (231) (Chapter 4.4), and interpreted as trivial (≤ 0.19), small (0.20 – 0.59), moderate (0.60 – 1.19), large (1.20 – 1.99), very large (2.00 – 3.99), and extremely large (≥ 4.00) (255). Mean changes were also interpreted as ratios relative to the SWC. The SWC was calculated as 0.2 × between-subject SD (253). Comparisons in pre- and post-intervention primary outcome variables and changes in outcome variables between the IG and CG were also assessed using independent sample t-tests or Mann-Whitney U tests, with effect sizes as outlined above. Furthermore, to link changes in knee joint loads and COD performance with changes in cutting kinetic and kinematic variables, Pearson's correlations (parametric data) or Spearman's correlations (non-parametric) were calculated with 95% CIs. Correlations were evaluated as follows: trivial (0.00 - 0.09), small (0.10 – 0.29), moderate (0.30 – 0.49), large (0.50 – 0.69), very large (0.70 – 0.89), nearly perfect (0.90 – 0.99), and perfect (1.00) (254). A correlation cut-off value of ≥ 0.40 was considered relevant according to Welch et al. (596) who also investigated the biomechanical determinants of cutting performance. Thus, correlations greater than this value are only reported in the subsequent tables. Statistical significance was defined $p \leq 0.05$ for all tests.

8.4 RESULTS

8.4.1 Pre-to-post changes in cutting variables

Pre-to-post changes in primary outcome variables for the IG and CG are presented in Tables 8.3, 8.5, 8.7, and 8.9, containing descriptives, *p* values, effect sizes, and mean differences. The following pre-to-post changes will be discussed by cutting task and primary outcome variable.

8.4.1.1 CUT45: Pre-to-post changes in performance variables

No significant differences were observed in performance outcome variables completion time, GCT, and exit velocity between groups pre-intervention ($p \geq 0.652$, $g \leq 0.17$). A non-significant and small difference was observed between groups pre-intervention for approach velocity ($p = 0.333$, $g = -0.38$).

Table 8.2. Two-way mixed ANOVAs for CUT45 performance and injury risk variables

		Time (effect)			Group (interaction)		
	Variable	***p* value**	**η2**	**power**	***p* value**	**η2**	**power**
Performance	**Completion time**	0.026*	0.184	0.626	0.003*	0.295	0.874
	GCT	0.205	0.063	0.24	0.010*	0.239	0.767
	Exit velocity	0.045*	0.152	0.529	0.001**	0.352	0.943
	Approach velocity	0.190	0.068	0.254	0.010*	0.239	0.768
	Velocity at FFC	0.063	0.131	0.464	0.002*	0.330	0.921
	Cut angle	<0.001**	0.457	0.993	0.020*	0.198	0.667
Multiplanar knee joint load	**peak KAM**	0.013*	0.223	0.732	0.116	0.096	0.346
	peak KIRM	0.379	0.031	0.138	0.575	0.013	0.085
	peak KFM	0.400	0.029	0.131	0.381	0.031	0.138
Injury risk angles	**peak KAA**	0.003	0.309	0.895	0.405	0.028	0.129
	IC KAA	<0.001**	0.547	1.000	0.267	0.049	0.194
	peak KRA	<0.001**	0.513	0.998	0.041*	0.156	0.543
	IC KRA	<0.001**	0.527	0.999	0.032*	0.172	0.590

Key: GCT: Ground contact time; FFC: Final foot contact; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; KFM: Knee flexion moment; KAA: Knee abduction angle; IC: Initial contact; KRA: Knee rotation angle; *: $p \leq 0.05$; **: $p \leq 0.001$

Trivial η2 (< 0.010)	**Small η2 (0.010-0.059)**	**Medium η2 (0.060-0.149)**	**Large η2 (≥ 0.150)**

Table 8.2 contains the results from the two-way mixed ANOVAs for performance variables, showing large and significant main effects of time for completion time, exit velocity, and cut angle. Large and significant interaction effect of time and group for all CUT45 performance variables were observed (Table 8.2), with the IG displaying faster completion times (p = 0.035, g = -0.83), shorter GCTs (p = 0.109, g = 0.83), greater exit velocities (p = 0.010, g = 1.03), greater approach velocities (p = 0.107, g = 0.62), and greater velocities at FFC (p = 0.003, g = 1.21) post-intervention compared to the CG. Pre-to-post changes in CUT45 performance variables are presented in Table 8.3 and Figure 8.2 for the IG and CG. Moderate and significant improvements in completion time, exit velocity, approach velocity, and velocity at FFC were observed for the IG only, while a small and significant improvement in GCT was also observed for the IG. Mean changes in completion time (p = 0.003, g = -1.24), GCT (p = 0.005, g = -1.08), exit velocity (p = 0.001, g = 1.38), approach velocity (p = 0.010, g = 1.07), and velocity at FFC (p = 0.002, g = 1.31) were significantly greater for the IG compared to the CG, with moderate to large effect sizes.

<u>8.4.1.2 CUT45: Pre-to-post changes in injury risk parameters</u>

No significant differences were observed in injury risk parameters (peak moments and angles) between groups pre-intervention ($p \geq 0.624$, $g \leq 0.19$).

Table 8.2 contains the results from the two-way mixed ANOVAs for injury risk variables, showing large and significant main effects of time for peak KAM, peak KAA, IC KAA, and peak KRA and IC KRA. Large and significant interaction effects of time and group were observed only for KRA variables, and a medium, yet non-significant interaction effect of time and group for CUT45 peak KAM was also observed, with the CG showing significantly greater peak KAMs (p = 0.013, g = -1.00) post-intervention compared to the IG.

Pre-to-post changes in CUT45 injury risk parameters are presented in Table 8.3 for the IG and CG. Small and non-significant increases in CUT45 peak KAMs were observed for the IG (Table 8.3, Figure 8.3) post-intervention, which were not greater than the SWC. It is worth noting that large individual variation for changes in peak KAMs were observed in the IG, with five participants showing reductions in peak KAMs greater than the SWC (Figure 8.3a) and SEM established in Chapter 4, and seven participants showing increases greater than the SWC and SEM. Importantly, the CG demonstrated a large and significant increase in peak KAM post-intervention (Table 8.3, Figure 8.3). Small and non-significant increases in CUT45 peak KIRMs were observed for the IG (Table 8.3, Figure 8.3) post-intervention, which was greater than the SWC and SEM (Chapter 4). It is worth noting that large individual variation for changes in peak KIRMs were observed in the IG, with five participants showing reductions in peak KIRMs greater than the SWC and SEM (Figure 3b), and seven participants showing increases greater than SWC and SEM. Trivial and non-significant changes in peak KIRMs were observed for the CG post-intervention. No significant time and interaction effect for KFMs were observed, and changes in peak KFMs were non-significant and trivial and small for the IG and CG, respectively. IC and peak KAAs significantly increased for the IG, while IC and peak KRA decreased post-intervention for the IG and CG (Table 8.3).

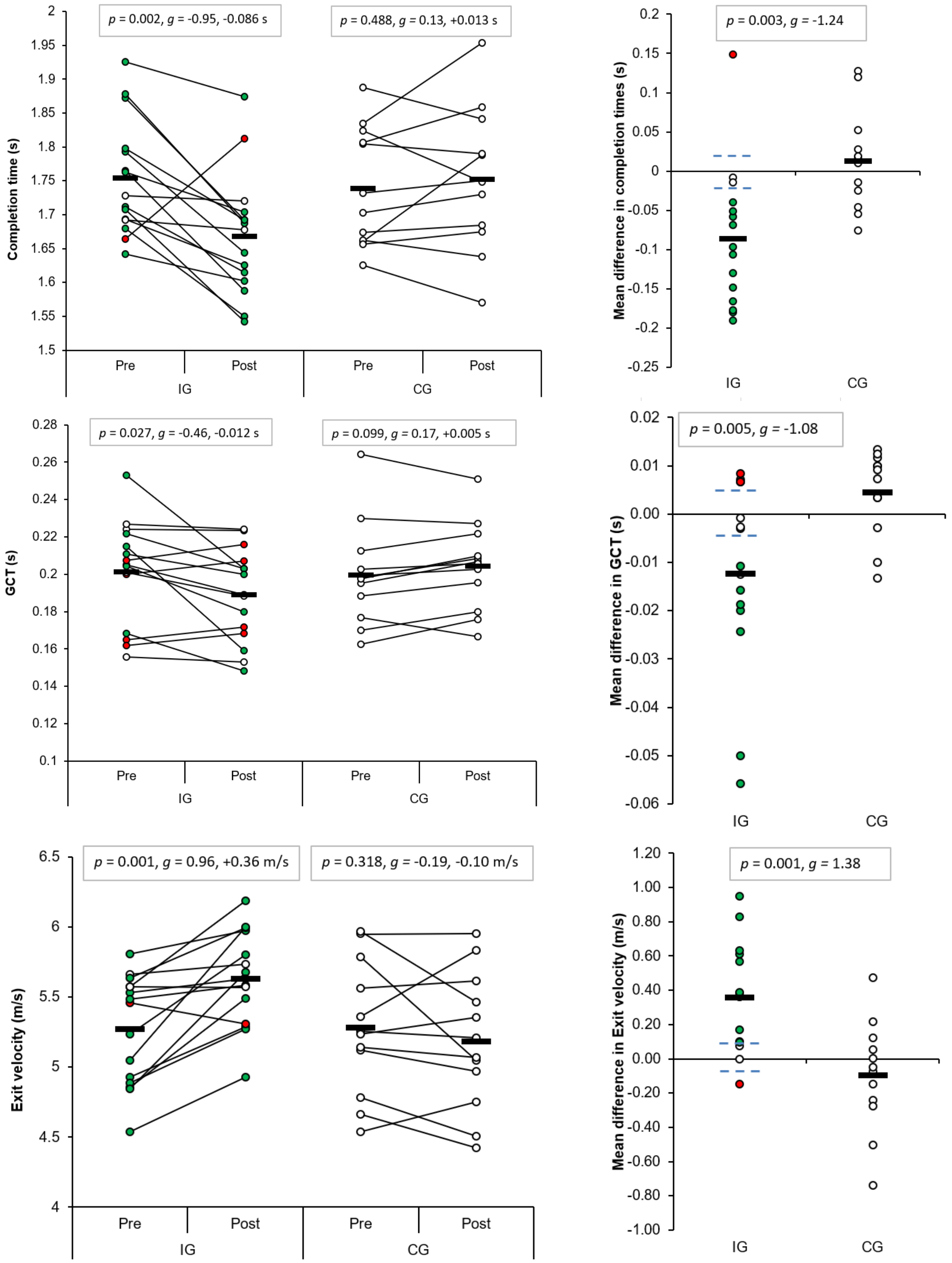


Figure 8.2. Pre-to-post changes in CUT45 performance variables. A: completion time; B: GCT; C: exit velocity. IG: Intervention group; CG: Control group; GCT: Ground contact time; green = positive responder; red = negative responder; white = non-responder; blue line = SWC.

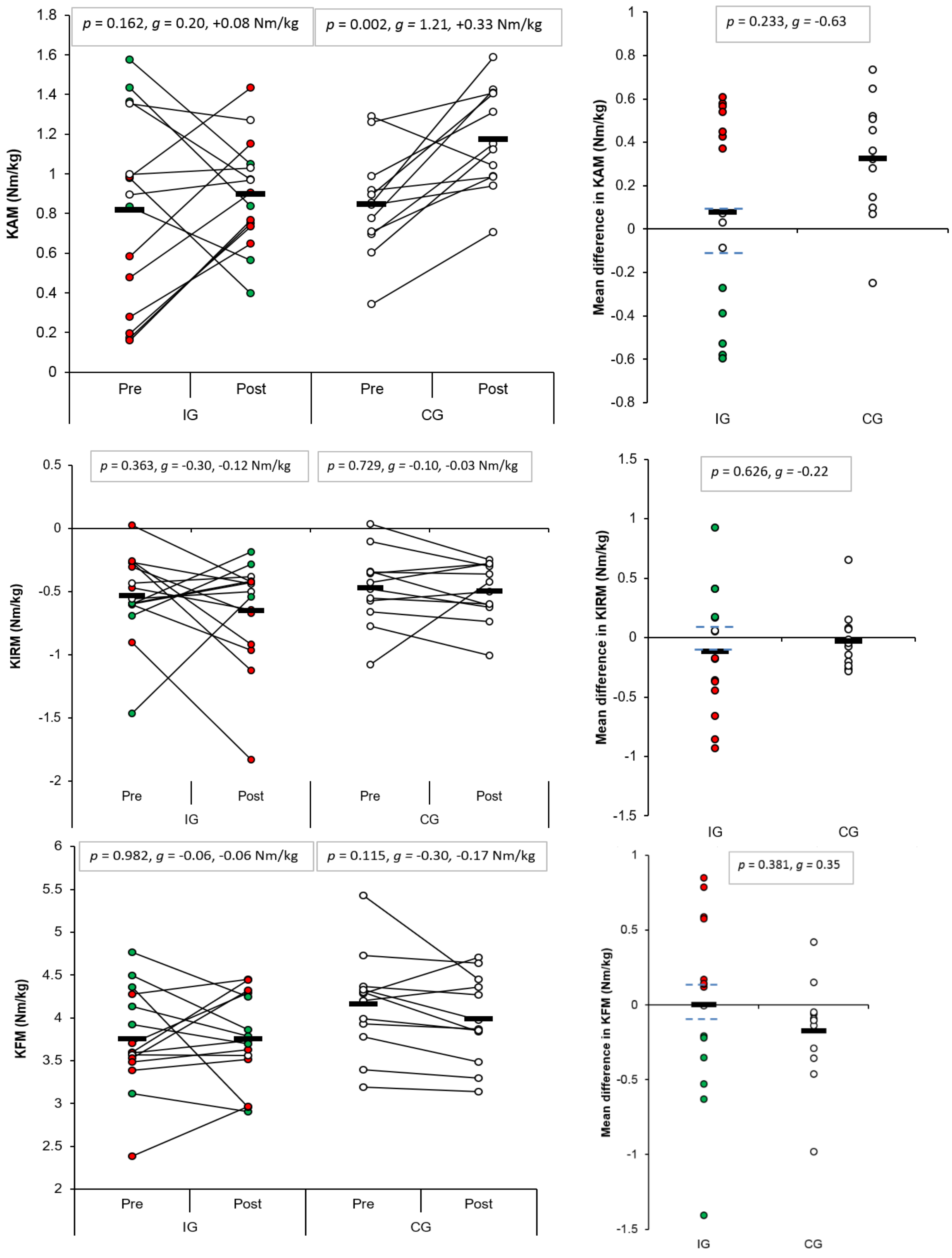


Figure 8.3. Pre-to-post changes in CUT45 multiplanar knee joint loads. A: KAM; B: KIRM; C: KFM. IG: Intervention group; CG: Control group; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; KFM: Knee flexion moment; green = positive responder; red = negative responder; white = non-responder; blue line = SWC.

Table 8.3. CUT45 Pre-to-post changes in performance and injury risk variables for IG and CG

	Group	Variable	Pre		Post		*p*	Hedges' *g* effect size		Mean difference		SWC	Ratio to SWC
			Mean	SD	Mean	SD		*g*	± CI	Mean	SD		
Performance	IG	**Completion time (s)**	1.754	0.085	1.668	0.091	0.002*	-0.95	0.76	-0.086	0.089	0.017	5.1
		GCT (s)	0.201	0.028	0.189	0.025	0.027*	-0.46	0.73	-0.012	0.020	0.006	2.2
		Exit velocity (m/s)	5.27	0.39	5.63	0.34	0.001**	0.96	0.76	0.36	0.32	0.08	4.6
		Approach velocity (m/s)	5.18	0.27	5.37	0.23	0.013*	0.74	0.74	0.19	0.26	0.05	3.6
		Velocity at FFC (m/s)	5.08	0.29	5.41	0.29	0.001**	1.10	0.77	0.33	0.29	0.06	5.6
		Cut angle (°)	28.8	2.7	29.5	2.5	0.281	0.27	0.72	0.7	1.9	0.5	1.3
	CG	**Completion time (s)**	1.739	0.088	1.752	0.104	0.488	0.13	0.80	0.013	0.064	0.018	0.8
		GCT (s)	0.200	0.027	0.204	0.023	0.099	0.17	0.80	0.005	0.009	0.005	0.8
		Exit velocity (m/s)	5.28	0.48	5.18	0.49	0.318	-0.19	0.80	-0.10	0.32	0.10	1.0
		Approach velocity (m/s)	5.27	0.22	5.21	0.27	0.288	-0.26	0.80	-0.07	0.21	0.04	1.5
		Velocity at FFC (m/s)	5.04	0.31	4.95	0.44	0.347	-0.24	0.80	-0.09	0.33	0.06	1.5
		Cut angle (°)	28.0	1.1	30.5	1.7	0.002*	1.64	0.93	2.4	1.6	0.2	11.0
Multiplanar knee joint load and injury risk angles	IG	**peak KAM (Nm/kg)**	0.82	0.49	0.90	0.27	0.162	0.20	0.72	0.08	0.46	0.10	0.8
		peak KIRM (Nm/kg)	-0.53	0.34	-0.65	0.42	0.363	-0.30	0.72	-0.12	0.51	0.07	1.7
		peak KFM (Nm/kg)	3.75	0.60	3.76	0.53	0.982	0.01	0.72	0.00	0.60	0.12	0.0
		peak KAA (°)	-9.4	5.5	-11.7	4.1	0.020*	-0.44	0.72	-2.2	4.5	1.1	2.0
		IC KAA (°)	2.7	5.4	-2.5	3.5	0.047*	-1.12	0.77	-5.2	4.7	1.1	4.8
		peak KRA (°)	-2.3	7.7	5.2	4.4	0.009*	1.16	0.77	7.5	5.9	1.5	4.8
		IC KRA (°)	-1.1	7.5	7.3	4.6	0.032*	1.32	0.79	8.5	6.3	1.5	5.6
	CG	**peak KAM (Nm/kg)**	0.85	0.26	1.17	0.26	0.002*	1.21	0.87	0.33	0.28	0.05	6.2
		peak KIRM (Nm/kg)	-0.47	0.29	-0.50	0.23	0.729	-0.10	0.80	-0.03	0.26	0.06	0.4
		peak KFM (Nm/kg)	4.16	0.59	3.99	0.51	0.115	-0.30	0.80	-0.17	0.34	0.12	1.4
		peak KAA (°)	-10.3	6.8	-14.1	6.3	0.018*	-0.54	0.81	-3.7	4.6	1.4	2.7
		IC KAA (°)	2.1	3.7	-1.3	3.7	0.003*	-0.90	0.84	-3.4	3.2	0.7	4.6
		peak KRA (°)	-3.1	5.7	-0.1	4.6	0.036*	0.58	0.73	3.1	4.4	1.1	2.7
		IC KRA (°)	-2.0	6.2	1.3	5.0	0.042*	0.58	0.73	3.4	5.1	1.2	2.7

Key: GCT: Ground contact time; FFC: Final foot contact; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; KFM: Knee flexion moment; KAA: Knee abduction angle; IC: Initial contact; KRA: Knee rotation angle; IG: Intervention group; CG: Control group; CI : 95% Confidence interval; SD: Standard deviation; SWC: Smallest worthwhile change; ES : Effect size; *: $p \leq 0.05$; **: $p \leq 0.001$

Trivial ES (≤ 0.19)	Small ES (0.20-0.59)	Moderate ES (0.60-1.19)	Large ES (1.20-1.99)

8.4.1.3 Pre-to-post changes in CUT45 technical and mechanical variables

Table 8.4 contains the results from the two-way mixed ANOVAs for technical and mechanical variables, showing large and significant main effects of time for peak and mean MLPF, IFPA, IC hip rotation angle, IC HFA, peak and IC KFA and ROM. Medium to large significant interaction effects of time and group were observed for peak and mean MLPF, and peak KFA and ROM. Pre-to-post changes in CUT45 technical and mechanical variables are presented in Table 8.5 for the IG and CG. Participants in the IG produced significantly greater peak and mean MLPFs, greater IFPAs, greater hip external rotation at IC, and smaller KFAs and ADFA ROMs post-intervention, with small to moderate effect sizes. Participants in the CG demonstrated significantly greater IFPAs post-intervention only. No other statistically significant changes in cutting mechanics were observed post-intervention for the IG or CG, such as peak VBF, lateral trunk flexion angle, and lateral foot plant distance (Table 8.5).

Table 8.4. Two-way mixed ANOVAs for CUT45 technical and mechanical variables

		Time (effect)			Group (interaction)		
	Variable	***p* value**	**η2**	**power**	***p* value**	**η2**	**power**
GRF	**peak MLPF**	0.009*	0.244	0.780	0.028*	0.179	0.613
	mean MLPF	0.002*	0.321	0.910	0.023*	0.189	0.642
	FFC peak VBF	0.392	0.030	0.134	0.973	0.000	0.050
Moment	**FFC peak HFM**	0.263	0.050	0.197	0.178	0.071	0.266
	FFC peak ADFM	0.698	0.006	0.066	0.442	0.024	0.117
Trunk, pelvis, hip, and foot	**Lateral trunk flexion – IC**	0.777	0.003	0.059	0.585	0.012	0.083
	FFC trunk inclination angle - IC	0.985	0.000	0.050	0.762	0.004	0.060
	Pelvic rotation – IC	0.269	0.049	0.193	0.821	0.002	0.056
	IFPA – IC	0.002*	0.335	0.927	0.396	0.029	0.132
	Lateral foot plant distance - IC	0.281	0.046	0.185	0.235	0.056	0.216
	HRA - IC	0.012*	0.227	0.740	0.183	0.070	0.261
	peak HRA	0.069	0.126	0.448	0.557	0.014	0.088
Sag hip, knee, and ankle angles	**FFC peak HFA**	0.065	0.130	0.460	0.903	0.001	0.052
	FFC IC HFA	0.034*	0.168	0.580	0.683	0.007	0.068
	FFC HFA ROM	0.285	0.046	0.183	0.104	0.102	0.368
	FFC peak KFA	<0.001**	0.994	1.000	0.046*	0.150	0.523
	FFC IC KFA	0.042*	0.155	0.540	0.349	0.035	0.151
	FFC KFA ROM	0.007*	0.260	0.814	0.049*	0.147	0.513
	FFC peak ADFA	0.740	0.004	0.062	0.316	0.040	0.166
	FFC IC ADFA	0.076	0.120	0.428	0.843	0.002	0.054
	FFC ADFA ROM	0.052	0.143	0.502	0.241	0.054	0.211

Key: MLPF: Medio-lateral propulsive force; BW: Body weight; VBF: Vertical braking force; HFM: Hip flexion moment; ADFM: Ankle dorsi-flexion moment; IFPA: Initial foot progression angle; HFA: Hip flexion angle; KFA: Knee flexion angle; ADFA: Ankle dorsi-flexion angle; HRA: Hip rotation angle; ROM: Range of motion; IC: Initial contact; FFC: Final foot contact; GRF: Ground reaction force; *: $p \leq 0.05$; **: $p \leq 0.001$

Trivial η2 (< 0.010)	**Small η2 (0.010-0.059)**	**Medium η2 (0.060-0.149)**	**Large η2 (≥ 0.150)**

Table 8.5. CUT45 Pre-to-post changes in cutting mechanical and technical variables for IG and CG

	Group	Variable	Pre		Post			Hedges' *g* effect size		Mean difference		SWC	Ratio to SWC
			Mean	SD	Mean	SD	*p*	*g*	± CI	Mean	SD		
GRF	IG	**peak MLPF (BW)**	1.38	0.19	1.50	0.23	0.005*	0.58	0.73	0.13	0.15	0.04	3.3
		mean MLPF (BW)	0.89	0.12	0.99	0.16	0.003*	0.69	0.74	0.10	0.11	0.02	4.2
		FFC peak VBF (BW)	4.02	0.95	4.14	1.08	0.566	0.11	0.72	0.12	0.77	0.19	0.6
	CG	**peak MLPF (BW)**	1.31	0.33	1.33	0.32	0.647	0.04	0.80	0.01	0.09	0.07	0.2
		mean MLPF (BW)	0.84	0.20	0.85	0.20	0.241	0.08	0.80	0.02	0.05	0.04	0.4
		FFC peak VBF (BW)	3.40	1.05	3.51	1.18	0.272	0.09	0.80	0.11	0.50	0.21	0.5
Moments	IG	**FFC peak HFM (Nm/kg)**	-3.90	1.24	-3.96	0.90	0.872	-0.06	0.72	-0.06	1.52	0.25	0.3
		FFC peak ADFM (Nm/kg)	-1.91	0.47	-2.03	0.64	0.446	-0.19	0.72	-0.11	0.56	0.09	1.2
	CG	**FFC peak HFM (Nm/kg)**	-4.34	1.24	-3.66	0.82	0.077	0.63	0.82	0.68	1.21	0.25	2.7
		FFC peak ADFM (Nm/kg)	-2.04	0.38	-2.00	0.39	0.756	0.09	0.80	0.04	0.41	0.08	0.5
Trunk, pelvis, hip, and foot	IG	**Lateral trunk flexion – IC (°)**	-21.9	6.2	-21.6	8.3	0.877	0.04	0.72	0.3	8.3	1.2	0.3
		FFC trunk inclination angle - IC (°)	4.5	2.8	4.7	3.5	1.000	0.05	0.72	0.1	3.1	0.6	0.3
		Pelvic rotation – IC (°)	4.4	7.0	5.8	7.0	0.417	0.20	0.72	1.4	6.6	1.4	1.0
		IFPA – IC (°)	-1.6	7.1	3.3	6.4	0.018*	0.70	0.74	4.8	7.0	1.4	3.4
		Lateral foot plant distance – IC (°)	-0.390	0.028	-0.403	0.029	0.160	-0.46	0.73	-0.013	0.035	0.006	2.4
		HRA - IC (°)	7.9	7.0	11.8	6.6	0.014*	0.56	0.73	4.0	5.4	1.4	2.8
		peak HRA (°)	5.3	7.6	7.9	7.2	0.094	0.34	0.72	2.6	5.6	1.5	1.7
	CG	**Lateral trunk flexion – IC (°)**	-19.6	9.2	-20.7	9.4	0.270	-0.11	0.80	-1.1	3.2	1.8	0.6
		FFC trunk inclination angle - IC (°)	5.0	4.0	4.8	4.1	1.000	-0.04	0.80	-0.2	2.0	0.8	0.2
		Pelvic rotation – IC (°)	4.8	6.6	5.8	5.4	0.347	0.15	0.80	1.0	3.4	1.3	0.7
		IFPA – IC (°)	-0.3	5.5	2.6	5.5	0.015*	0.52	0.81	3.0	3.2	1.1	2.7
		Lateral foot plant distance - IC (°)	-0.370	0.045	-0.370	0.004	0.917	0.01	0.80	0.001	0.022	0.009	0.1
		HRA - IC (°)	8.0	3.6	9.3	6.3	0.330	0.24	0.80	1.3	4.4	0.7	1.8
		peak HRA (°)	5.9	3.3	7.2	6.5	0.372	0.25	0.80	1.4	5.0	0.7	2.0
Sag hip, knee, and ankle angles	IG	**FFC peak HFA (°)**	54.8	9.0	56.7	6.5	0.262	0.24	0.72	1.9	6.4	1.8	1.1
		FFC IC HFA (°)	53.4	9.2	56.3	6.5	0.094	0.35	0.72	2.9	6.2	1.8	1.6
		FFC HFA ROM (°)	1.3	2.3	0.4	0.8	0.091	-0.54	0.73	-1.0	2.1	0.5	2.1
		FFC peak KFA (°)	54.8	6.1	52.2	4.6	0.066	-0.46	0.73	-2.6	5.0	1.2	2.1
		FFC IC KFA (°)	27.4	6.4	29.8	4.4	0.063	0.44	0.72	2.5	4.7	1.3	1.9
		FFC KFA ROM (°)	27.4	8.9	22.4	7.7	0.004*	-0.59	0.73	-5.0	5.8	1.8	2.8
		FFC peak ADFA (°)	77.2	6.4	76.0	4.7	0.383	-0.22	0.72	-1.3	5.7	1.3	1.0
		FFC IC ADFA (°)	56.6	6.1	58.4	4.9	0.179	0.31	0.72	1.8	4.9	1.2	1.5
		FFC ADFA ROM (°)	20.6	5.6	17.6	3.5	0.017*	-0.64	0.73	-3.0	4.3	1.1	2.7
	CG	**FFC peak HFA (°)**	53.1	6.9	55.3	8.6	0.092	0.27	0.80	2.2	4.1	1.4	1.6
		FFC IC HFA (°)	52.6	7.0	54.6	8.9	0.171	0.24	0.80	2.0	4.7	1.4	1.4
		FFC HFA ROM (°)	0.5	1.0	0.8	1.2	0.674	0.19	0.80	0.2	1.2	0.2	1.1
		FFC peak KFA (°)	57.1	4.6	57.1	4.1	0.960	0.01	0.80	0.1	3.7	0.9	0.1
		FFC IC KFA (°)	27.4	6.8	28.4	5.9	0.319	0.14	0.80	0.9	3.1	1.4	0.7
		FFC KFA ROM (°)	29.7	7.7	28.8	6.2	0.483	-0.12	0.80	-0.9	4.2	1.5	0.6
		FFC peak ADFA (°)	78.6	3.5	79.3	5.5	0.529	0.13	0.80	0.6	3.4	0.7	0.9
		FFC IC ADFA (°)	58.7	5.7	60.1	5.1	0.233	0.26	0.80	1.4	3.9	1.1	1.2
		FFC ADFA ROM (°)	20.0	5.5	19.2	5.3	0.623	-0.14	0.80	-0.8	5.4	1.1	0.7

Key: MLPF: Medio-lateral propulsive force; BW: Body weight; VBF: Vertical braking force; HFM: Hip flexion moment; ADFM: Ankle dorsi-flexion moment; IFPA: Initial foot progression angle; HFA: Hip flexion angle; KFA: Knee flexion angle; ADFA: Ankle dorsi-flexion angle; HRA: Hip rotation angle; ROM: Range of motion; IC: Initial contact; FFC: Final foot contact; IG: Intervention group; CG: Control group; CI : 95% Confidence interval; SD: Standard deviation; SWC: Smallest worthwhile change; ES : Effect size; GRF: Ground reaction force; *: $p \leq 0.05$; **: $p \leq 0.001$

Trivial ES (≤ 0.19)	Small ES (0.20-0.59)	Moderate ES (0.60-1.19)	Large ES (1.20-1.99)

8.4.2.1 CUT90: Pre-to-post changes in performance variables

No significant differences were observed in CUT90 performance outcome variables completion time, GCT, and exit velocity between groups pre-intervention ($p \geq 0.652$, $g \leq 0.28$). A non-significant and small difference was observed between groups pre-intervention for approach velocity ($p = 0.333$, $g = -0.57$).

Table 8.6. Two-way mixed ANOVAs for CUT90 performance and injury risk variables

		Time (effect)			Group (interaction)		
	Variable	***p* value**	**η2**	**power**	***p* value**	**η2**	**power**
Performance	**Completion time**	<0.001**	0.539	0.999	<0.001**	0.483	0.996
	GCT	0.768	0.004	0.059	0.272	0.048	0.191
	Exit velocity	0.249	0.053	0.206	0.051	0.144	0.506
	Approach velocity	0.263	0.050	0.196	0.586	0.012	0.083
	Velocity at FFC	0.535	0.016	0.093	0.470	0.021	0.109
	Cut angle	0.327	0.038	0.161	0.715	0.005	0.065
Multiplanar knee joint load	**peak KAM**	0.488	0.019	0.104	0.124	0.092	0.334
	peak KIRM	0.922	0.000	0.051	0.292	0.044	0.179
	peak KFM	0.594	0.012	0.081	0.998	0.000	0.050
Injury risk angles	**peak KAA**	0.596	0.011	0.081	0.099	0.105	0.377
	IC KAA	0.887	0.001	0.052	0.290	0.045	0.180
	peak KRA	0.008*	0.252	0.796	0.418	0.026	0.125
	IC KRA	<0.001**	0.511	0.998	0.180	0.071	0.264

Key: GCT: Ground contact time; FFC: Final foot contact; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; KFM: Knee flexion moment; KAA: Knee abduction angle; IC: Initial contact; KRA: Knee rotation angle; *: $p \leq 0.05$; **: $p \leq 0.001$

Trivial η2 (< 0.010)	**Small η2 (0.010-0.059)**	**Medium η2 (0.060-0.149)**	**Large η2 (≥ 0.150)**

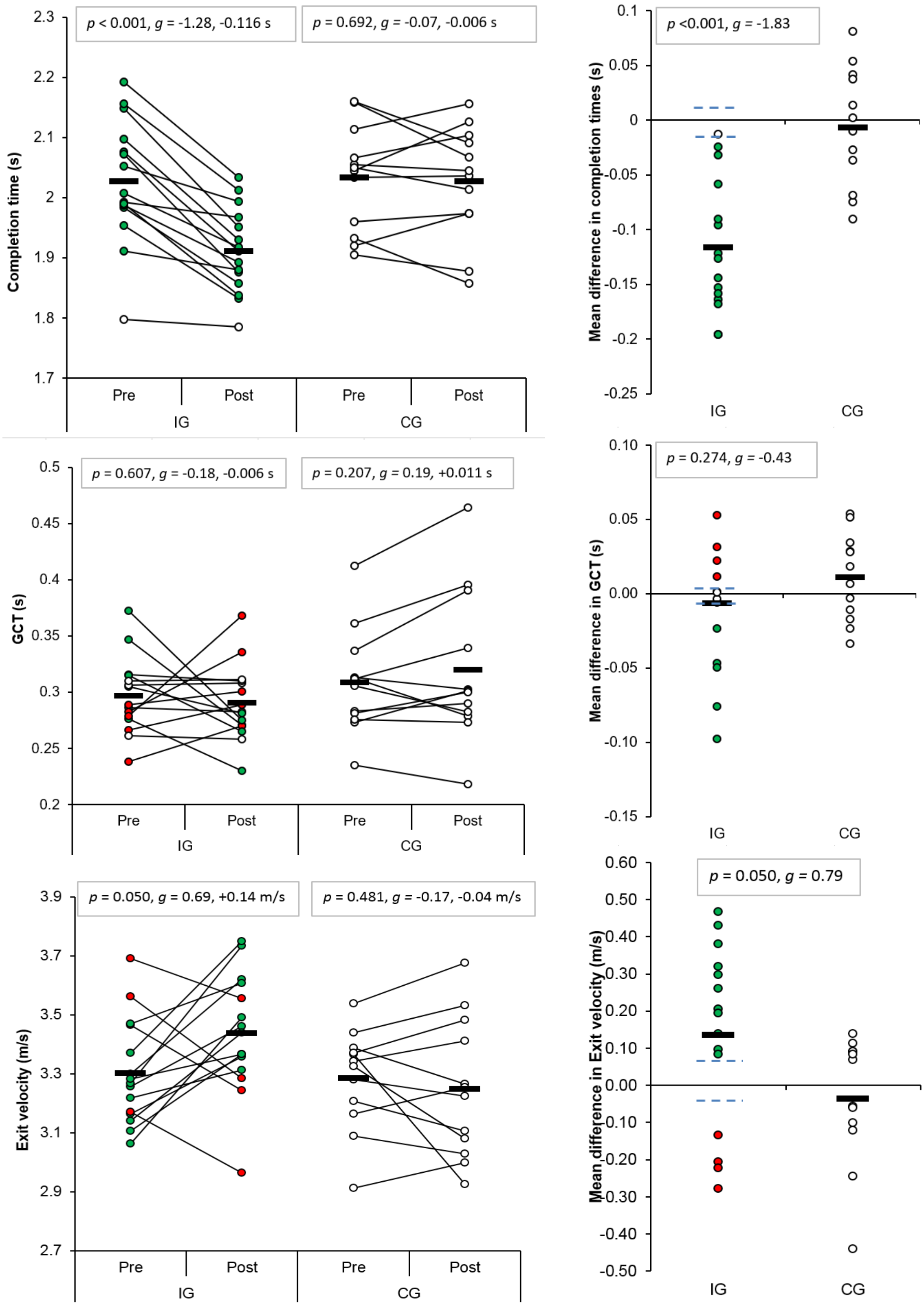


Figure 8.4. Pre-to-post changes in CUT90 performance variables. A: Completion time; B: GCT; C: Exit velocity. IG: Intervention group; CG: Control group; GCT: Ground contact time; green = positive responder; red = negative responder; white = non-responder; blue line = SWC.

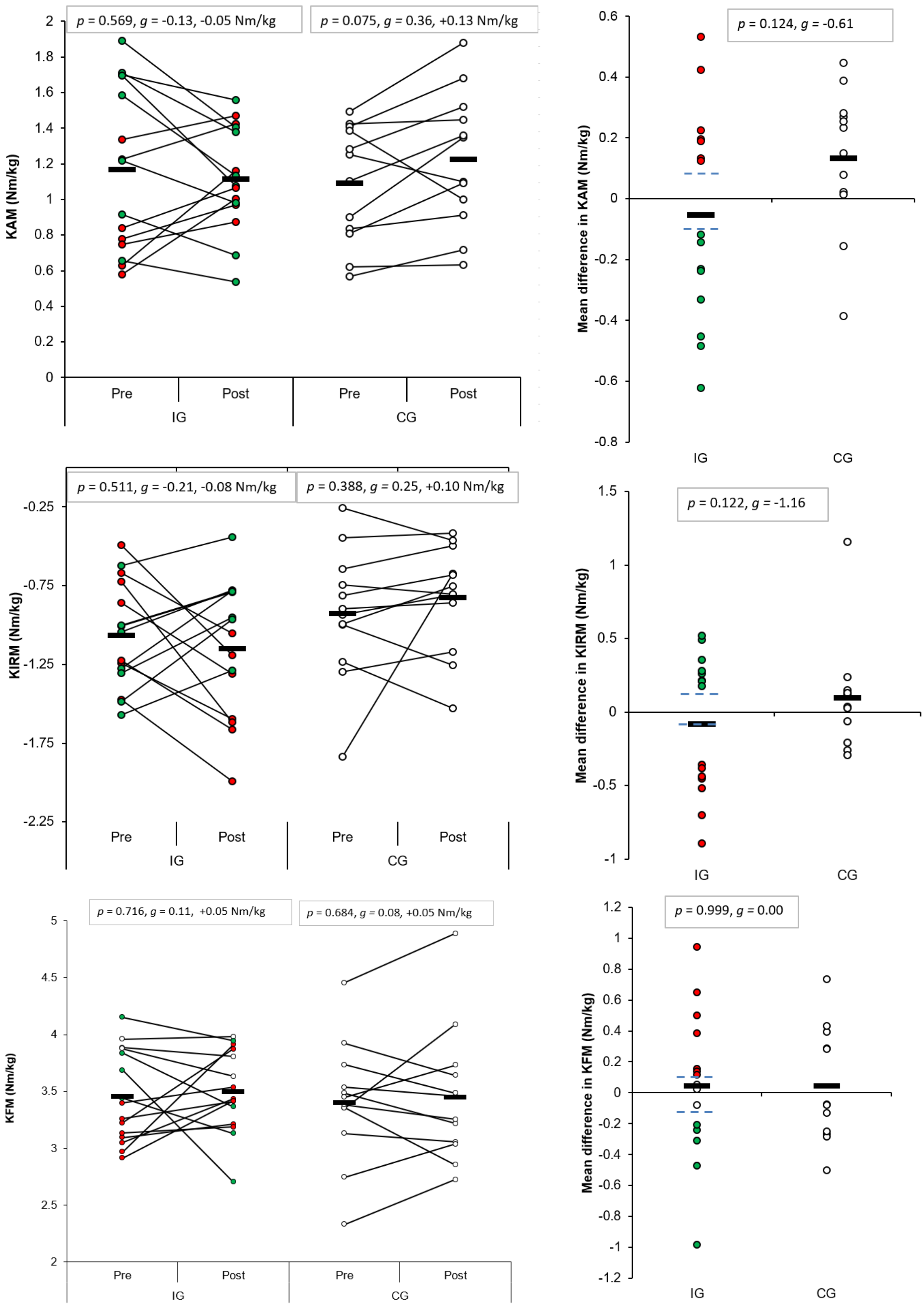


Figure 8.5. Pre-to-post changes in CUT90 injury risk multiplanar knee joint loads. A: KAM; B: KIRM; C: KFM. IG: Intervention group; CG: Control group; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; KFM: Knee flexion moment; green = positive responder; red = negative responder; white = non-responder; blue line = SWC.

Table 8.6 contains the results from the two-way mixed ANOVAs for CUT90 performance variables, showing large and significant main effects of time for completion time only. A large and significant interaction effect of time and group for CUT90 completion time was observed, with the IG showing significantly faster post-intervention completion times (p = 0.031, g = -1.35) compared to the CG. Additionally, a medium and non-significant interaction effect was observed for exit velocity (Table 8.6), with the IG showing significantly greater exit velocities (p = 0.035, g = 0.83) post-intervention compared to the CG.

Pre-to-post changes in CUT90 performance variables are presented in Table 8.7 for the IG and CG. A large and moderate significant improvement was observed for completion time and exit velocity (Table 8.5 and Figure 8.4), respectively, for the IG only. Mean changes in completion time (p < 0.001, g = -1.83), and exit velocity (p = 0.050, g = 0.79) were significantly greater for the IG compared to the CG, with moderate to large effect sizes.

8.4.2.2 CUT90: Pre-to-post changes in injury risk variables

No significant differences were observed in injury risk variables (peak moments and angles) between groups pre-intervention ($p \geq 0.335$, g = 0.04-0.48).

Table 8.6 contains the results from the two-way mixed ANOVAs for injury risk variables, showing large and significant main effects of time for peak KRA and IC KRA only. No statistically significant interaction effects were observed for injury risk variables. Pre-to-post changes in CUT90 injury risk variables are presented in Table 8.7 for the IG and CG. Changes in peak KAMs were non-significant and trivial for the IG (Table 8.7, Figure 8.5a) post-intervention, which were not greater than the SWC and SEM. It is worth noting that large individual variation for changes in peak KAMs were observed in the IG, with eight participants showing reductions in peak KAMs greater than the SWC and SEM (Figure 8.5a), and seven participants showing increases greater than SWC and SEM established in Chapter 4. The CG demonstrated a small and non-significant increase in peak KAM post-intervention. A small and non-significant increase in CUT90 peak KIRM was observed for the IG (Table 8.7, Figure 8.5b) post-intervention, which was not greater than the SWC and SEM. It is worth noting that large individual variation in changes in peak KIRM were observed in the IG, with eight participants showing reductions in peak KIRMs greater than the SWC and SEM (Figure 8.5b), and seven participants showing increases greater than SWC and SEM. A small and non-significant reduction in peak KIRMs were observed for the CG post-intervention. No significant time and interaction effect for KFMs was observed, and changes in peak KFMs were non-significant and trivial for the IG and CG. IC and peak

KAAs increased post-intervention for both groups, while IC and peak KRAs were significantly reduced for the IG post-intervention (Table 8.7), with a moderate effect size.

Table 8.7. CUT90 Pre-to-post changes in performance and injury risk variables for IG and CG

	Group	Variable	Pre		Post			Hedges' *g* effect size		Mean difference		SWC	Ratio to SWC
			Mean	SD	Mean	SD	*p*	*g*	± CI	Mean	SD		
Performance	IG	Completion time (s)	2.028	0.102	1.912	0.071	<0.001**	-1.28	0.79	-0.116	0.061	0.020	5.7
		GCT (s)	0.297	0.034	0.291	0.034	0.607	-0.18	0.72	-0.006	0.048	0.007	0.9
		Exit velocity (m/s)	3.30	0.18	3.44	0.20	0.050*	0.69	0.74	0.14	0.24	0.04	3.8
		Approach velocity (m/s)	4.43	0.33	4.51	0.24	0.156	0.29	0.72	0.09	0.31	0.07	1.3
		Velocity at FFC (m/s)	3.41	0.27	3.47	0.32	0.396	0.18	0.72	0.05	0.24	0.05	1.0
		Cut angle (°)	54.3	2.6	55.1	2.5	0.211	0.31	0.72	0.8	3.8	0.5	1.6
	CG	Completion time (s)	2.033	0.088	2.027	0.093	0.692	-0.07	0.80	-0.006	0.055	0.018	0.4
		GCT (s)	0.309	0.046	0.320	0.067	0.207	0.19	0.80	0.011	0.029	0.009	1.2
		Exit velocity (m/s)	3.29	0.17	3.25	0.24	0.481	-0.17	0.80	-0.04	0.17	0.03	1.1
		Approach velocity (m/s)	4.61	0.28	4.64	0.29	0.550	0.10	0.80	0.03	0.17	0.06	0.5
		Velocity at FFC (m/s)	3.46	0.25	3.46	0.29	0.925	-0.02	0.80	0.00	0.15	0.05	0.1
		Cut angle (°)	52.8	4.5	53.2	4.7	0.136	0.08	0.80	0.4	2.0	0.9	0.4
Multiplanar knee joint load and injury risk angles	IG	peak KAM (Nm/kg)	1.17	0.46	1.11	0.29	0.569	-0.13	0.72	-0.05	0.34	0.09	0.6
		peak KIRM (Nm/kg)	-1.07	0.34	-1.15	0.43	0.511	-0.21	0.72	-0.08	0.47	0.07	1.2
		peak KFM (Nm/kg)	3.46	0.41	3.50	0.36	0.716	0.11	0.72	0.05	0.47	0.08	0.6
		peak KAA (°)	-11.9	6.0	-15.6	5.2	0.023*	-0.64	0.73	-3.7	4.8	1.2	3.1
		IC KAA (°)	3.8	3.8	0.1	4.5	<0.001**	-0.86	0.75	-3.7	2.2	0.8	4.9
		peak KRA (°)	-7.5	7.2	-1.6	5.9	<0.001**	0.88	0.75	6.0	3.6	1.4	4.1
		IC KRA (°)	-4.8	7.8	1.4	7.2	0.001**	0.81	0.74	6.2	5.4	1.6	4.0
	CG	peak KAM (Nm/kg)	1.09	0.33	1.22	0.38	0.075	0.36	0.81	0.13	0.24	0.07	2.0
		peak KIRM (Nm/kg)	-0.92	0.41	-0.83	0.34	0.388	0.25	0.80	0.10	0.38	0.08	1.2
		peak KFM (Nm/kg)	3.41	0.54	3.45	0.60	0.684	0.08	0.80	0.05	0.37	0.11	0.4
		peak KAA (°)	-13.2	8.0	-15.3	6.8	0.223	-0.27	0.80	-2.1	5.5	1.6	1.3
		IC KAA (°)	2.9	3.6	0.8	4.6	0.070	-0.50	0.81	-2.1	3.7	0.7	2.9
		peak KRA (°)	-6.0	7.4	-3.9	6.1	0.433	0.30	0.80	2.1	8.9	1.5	1.4
		IC KRA (°)	-5.1	7.0	-2.4	7.4	0.322	0.36	0.81	2.7	9.1	1.4	1.9

Key: GCT: Ground contact time; FFC: Final foot contact; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; KFM: Knee flexion moment; KAA: Knee abduction angle; IC: Initial contact; KRA: Knee rotation angle; IG: Intervention group; CG: Control group; CI : 95% Confidence interval; SD: Standard deviation; SWC: Smallest worthwhile change; ES: Effect size; *: $p \leq 0.05$; **: $p \leq 0.001$

Trivial ES (≤ 0.19)	Small ES (0.20-0.59)	Moderate ES (0.60-1.19)	Large ES (1.20-1.99)

8.4.2.3 Pre-to-post changes in CUT90 technical and mechanical variables data

Table 8.8. Two-way mixed ANOVAs for CUT90 technical and mechanical variables

		Time (effect)			Group (interaction)		
	Variable	*p* value	η2	power	*p* value	η2	power
GRF	peak MLPF	<0.001**	0.601	1.000	0.063	0.132	0.465
	mean MLPF	0.027*	0.181	0.619	0.009*	0.243	0.777
	peak HPF	0.083	0.115	0.411	0.016*	0.210	0.699
	mean HPF	0.007*	0.255	0.802	0.004*	0.290	0.866
	FFC mean HBF	0.002*	0.311	0.898	0.038*	0.160	0.556
	PFC mean HBF	0.036*	0.164	0.568	0.124	0.092	0.335
	FFC peak VBF	0.816	0.002	0.056	0.534	0.016	0.093
Moments	FFC peak HFM	0.181	0.070	0.263	0.148	0.082	0.301
	FFC peak ADFM	0.086	0.113	0.404	0.009*	0.241	0.772
Trunk, pelvis, hip, and foot	Lateral trunk flexion – IC	0.551	0.014	0.090	0.880	0.001	0.052
	FFC trunk inclination angle - IC	<0.001**	0.396	0.973	0.649	0.008	0.073
	Pelvic rotation – IC	0.073	0.123	0.437	0.068	0.127	0.449
	IFPA – IC	0.005*	0.274	0.838	0.007*	0.258	0.810
	Lateral foot plant distance - IC	0.585	0.012	0.083	0.837	0.002	0.055
	HRA - IC	0.274	0.048	0.190	0.833	0.002	0.055
	peak HRA	0.369	0.032	0.142	0.752	0.004	0.061
Sag hip, knee, and ankle angles	FFC peak HFA	0.178	0.071	0.266	0.492	0.019	0.103
	FFC IC HFA	0.077	0.120	0.427	0.660	0.008	0.071
	FFC HFA ROM	0.622	0.010	0.077	0.451	0.023	0.114
	FFC peak KFA	0.615	0.010	0.078	0.681	0.007	0.069
	FFC IC KFA	0.011*	0.233	0.754	0.360	0.034	0.146
	FFC KFA ROM	<0.001**	0.977	1.000	0.045*	0.151	0.528
	FFC peak ADFA	<0.001**	0.995	1.000	0.205	0.063	0.240
	FFC IC ADFA	0.907	0.001	0.051	0.077	0.120	0.425
	FFC ADFA ROM	0.045*	0.152	0.529	0.035*	0.166	0.572

Key: MLPF: Medio-lateral propulsive force; BW: Body weight; VBF: Vertical braking force; HFM: Hip flexion moment; ADFM: Ankle dorsi-flexion moment; IFPA: Initial foot progression angle; HFA: Hip flexion angle; KFA: Knee flexion angle; ADFA: Ankle dorsi-flexion angle; HRA: Hip rotation angle; ROM: Range of motion; IC: Initial contact; FFC: Final foot contact; GRF: Ground reaction force; *: $p \leq 0.05$; **: $p \leq 0.001$

Trivial η2 (< 0.010)	Small η2 (0.010-0.059)	Medium η2 (0.060-0.149)	Large η2 (≥ 0.150)

Table 8.9. CUT90 Pre-to-post changes in cutting technical and mechanical variables for IG and CG

	Group	Variable	Pre		Post			Hedges' *g* effect size		Mean difference		SWC	Ratio to SWC
			Mean	SD	Mean	SD	*p*	*g*	± CI	Mean	SD		
GRF	IG	peak MLPF (BW)	1.15	0.20	1.25	0.23	0.007*	0.44	0.72	0.10	0.12	0.04	2.5
		mean MLPF (BW)	0.82	0.16	0.90	0.15	0.003*	0.48	0.73	0.08	0.08	0.03	2.5
		peak HPF (BW)	-0.86	0.14	-0.97	0.20	0.003*	-0.65	0.73	-0.11	0.12	0.03	4.1
		mean HPF (BW)	-0.40	0.09	-0.48	0.09	<0.001**	-0.85	0.75	-0.08	0.06	0.02	4.4
		FFC mean HBF (BW)	-0.96	0.17	-1.07	0.18	0.006*	-0.60	0.73	-0.11	0.12	0.03	3.1
		PFC mean HBF (BW)	-0.59	0.13	-0.65	0.16	0.024*	-0.39	0.72	-0.06	0.09	0.03	2.2
		FFC peak VBF (BW)	3.03	0.87	3.10	0.81	0.307	0.07	0.72	0.06	0.32	0.17	0.4
	CG	peak MLPF (BW)	0.87	0.29	1.06	0.30	<0.001**	0.62	0.82	0.19	0.12	0.06	3.3
		mean MLPF (BW)	0.76	0.18	0.75	0.19	0.726	-0.04	0.80	-0.01	0.07	0.04	0.2
		peak HPF (BW)	-0.78	0.19	-0.76	0.23	0.638	0.09	0.80	0.02	0.14	0.04	0.5
		mean HPF (BW)	-0.36	0.11	-0.36	0.12	0.879	0.03	0.80	0.00	0.08	0.02	0.2
		FFC mean HBF (BW)	-0.88	0.22	-0.90	0.18	0.291	-0.11	0.80	-0.02	0.07	0.04	0.5
		PFC mean HBF (BW)	-0.57	0.15	-0.58	0.11	0.616	-0.07	0.80	-0.01	0.06	0.03	0.3
		FFC peak VBF (BW)	2.69	1.09	2.66	0.87	0.530	-0.03	0.80	-0.03	0.41	0.22	0.1
Moments	IG	FFC peak HFM (Nm/kg)	-3.29	0.93	-3.31	0.91	0.921	-0.02	0.72	-0.02	0.74	0.19	0.1
		FFC peak ADFM (Nm/kg)	-1.61	0.33	-1.89	0.38	0.009*	-0.78	0.74	-0.29	0.37	0.07	4.3
	CG	FFC peak HFM (Nm/kg)	-3.16	0.68	-2.69	0.63	0.118	0.69	0.82	0.47	0.96	0.14	3.5
		FFC peak ADFM (Nm/kg)	-1.84	0.36	-1.77	0.33	0.405	0.18	0.80	0.06	0.26	0.07	0.9
Trunk, pelvis, hip, and foot	IG	Lateral trunk flexion – IC (°)	-20.5	8.7	-21.1	8.0	0.768	-0.07	0.72	-0.6	8.3	1.7	0.4
		FFC trunk inclination angle - IC (°)	15.1	6.0	18.2	8.0	0.006*	0.43	0.72	3.1	3.7	1.2	2.6
		Pelvic rotation – IC (°)	28.6	9.9	33.7	9.5	0.032*	0.52	0.73	5.2	8.4	2.0	2.6
		IFPA – IC (°)	11.1	9.3	18.7	5.6	0.002*	0.97	0.76	7.6	7.7	1.9	4.1
		Lateral foot plant distance - IC (°)	-0.320	0.035	-0.318	0.038	0.822	0.07	0.72	0.003	0.048	0.007	0.4
		HRA - IC (°)	11.8	8.8	13.1	7.0	0.302	0.16	0.72	1.3	4.6	1.8	0.7
		peak HRA (°)	10.3	8.1	11.9	6.8	0.183	0.21	0.72	1.6	4.4	1.6	1.0
	CG	Lateral trunk flexion – IC (°)	-14.8	8.6	-15.9	7.2	0.544	-0.13	0.80	-1.1	6.0	1.7	0.6
		FFC trunk inclination angle - IC (°)	11.0	5.2	13.5	7.7	0.027*	0.36	0.81	2.5	3.4	1.0	2.4
		Pelvic rotation – IC (°)	33.5	12.4	33.4	12.7	0.976	0.00	0.80	0.0	4.9	2.5	0.0
		IFPA – IC (°)	16.7	11.9	16.8	13.6	0.917	0.01	0.80	0.1	4.8	2.4	0.1
		Lateral foot plant distance - IC (°)	-0.289	0.054	-0.282	0.073	0.541	0.10	0.80	0.006	0.035	0.011	0.6
		HRA - IC (°)	11.6	7.4	13.5	9.7	0.515	0.21	0.80	1.9	9.7	1.5	1.3
		peak HRA (°)	9.1	7.0	9.9	9.2	0.766	0.09	0.80	0.8	8.8	1.4	0.6
Sag hip, knee, and ankle angles	IG	FFC peak HFA (°)	51.7	7.3	52.8	6.4	0.609	0.14	0.72	1.0	7.8	1.5	0.7
		FFC IC HFA (°)	49.2	6.6	51.0	5.6	0.307	0.28	0.72	1.8	7.2	1.3	1.3
		FFC HFA ROM (°)	2.5	2.7	1.7	1.7	0.394	-0.33	0.72	-0.8	3.0	0.5	1.4
		FFC peak KFA (°)	62.7	5.9	61.7	5.2	0.518	-0.17	0.72	-1.0	5.7	1.2	0.8
		FFC IC KFA (°)	23.8	5.4	26.6	4.4	0.029*	0.54	0.73	2.8	4.4	1.1	2.5
		FFC KFA ROM (°)	38.9	6.6	35.2	5.7	0.018*	-0.59	0.73	-3.7	5.4	1.3	2.8
		FFC peak ADFA (°)	75.3	4.6	79.3	5.5	0.001**	0.77	0.74	4.0	3.8	0.9	4.3
		FFC IC ADFA (°)	50.8	7.1	49.8	7.3	0.255	-0.14	0.72	-1.0	3.4	1.4	0.7
		FFC ADFA ROM (°)	24.5	6.9	29.6	7.9	0.011*	0.67	0.74	5.1	6.2	1.4	3.7
	CG	FFC peak HFA (°)	50.3	7.8	53.4	10.2	0.173	0.33	0.81	3.1	7.3	1.6	2.0
		FFC IC HFA (°)	45.0	6.2	47.9	7.6	0.096	0.40	0.81	2.9	5.5	1.2	2.3
		FFC HFA ROM (°)	5.3	3.3	5.5	4.3	0.866	0.04	0.80	0.2	3.2	0.7	0.2
		FFC peak KFA (°)	66.0	5.1	65.9	6.5	0.948	-0.02	0.80	-0.1	5.1	1.0	0.1
		FFC IC KFA (°)	23.1	4.6	24.5	5.2	0.152	0.27	0.80	1.4	3.1	0.9	1.5
		FFC KFA ROM (°)	42.9	6.8	41.4	8.3	0.396	-0.19	0.80	-1.5	5.7	1.4	1.1
		FFC peak ADFA (°)	79.7	6.7	80.8	7.8	0.369	0.14	0.80	1.1	3.9	1.3	0.8
		FFC IC ADFA (°)	52.6	7.4	53.7	6.5	0.160	0.16	0.80	1.2	2.7	1.5	0.8
		FFC ADFA ROM (°)	27.2	10.2	27.0	9.7	0.939	-0.01	0.80	-0.1	5.8	2.0	0.1

Key: MLPF: Medio-lateral propulsive force; BW: Body weight; VBF: Vertical braking force; HFM: Hip flexion moment; ADFM: Ankle dorsi-flexion moment; IFPA: Initial foot progression angle; HFA: Hip flexion angle; KFA: Knee flexion angle; ADFA: Ankle dorsi-flexion angle; HRA: Hip rotation angle; HPF: Horizontal propulsive force; ROM: Range of motion; IC: Initial contact; FFC: Final foot contact; IG: Intervention group; CG: Control group; CI: 95% Confidence interval; SD: Standard deviation; SWC: Smallest worthwhile change; ES: Effect size; GRF: Ground reaction force; *: $p \leq 0.05$; **: $p \leq 0.001$

Trivial ES (≤ 0.19)	Small ES (0.20-0.59)	Moderate ES (0.60-1.19)	Large ES (1.20-1.99)

Table 8.8 contains the results from the two-way mixed ANOVAs for CUT90 technical and mechanical variables, showing large and significant effects of time for peak and mean MLPF, mean HPF, FFC and PFC mean HBF, IFPA, FFC trunk inclination angle, KFA at IC and ROM, and ADFA ROM. Medium to large significant interaction effects of time and group were observed for mean MLPF, peak and mean HPF, FFC mean HBF, FFC peak ADFM, IFPA, and knee and ADFA ROM (Table 8.8)

Pre-to-post changes in CUT90 technical and mechanical variables are presented in Table 8.9 for the IG and CG. Participants in the IG produced significantly greater peak ADFMs, greater peak and mean MLPFs, greater peak HPFs, greater mean HPFs, greater FFC mean HBFs, and greater PFC mean HBFs, with small to moderate effect sizes. Additionally, participants in the IG demonstrated significantly greater forward trunk inclination angles, greater pelvic rotation angles, greater IFPAs, greater initial KFAs, greater peak ADFAs, smaller KFA ROM, and greater ADFA ROM post-intervention, with small to moderate effect sizes. Participants in the CG produced significantly greater peak MLPFs, and greater forward trunk inclination angles. No other statistically significant changes in cutting mechanics were observed post-intervention for the IG or CG, such as peak VBF, lateral trunk flexion angle, and lateral foot plant distance (Table 8.9).

8.5.3 Associations between changes in performance and injury risk primary outcome variables with cutting mechanical variables

Table 8.10 present the correlation values between changes in performance and injury risk variables with changes in cutting technical and mechanical variables.

8.5.3.1 CUT45 correlation values between changes in performance and injury risk variables with cutting mechanical variables

- Improvements in completion time were very largely associated with increased exit velocity; largely associated with decreased GCT, increased IC HFA and external HRA; and moderately associated with decreased KFA ROM, mean MLPF, and increased FFC velocity (Table 8.10).
- Improvements in FFC GCT were very largely associated with decreased peak KFA and ROM; and largely associated increased approach and exit velocity, and decreased peak ADFA (Table 8.10).
- Improvements in exit velocity were very largely associated with increased velocity at FFC; largely associated with increased IC HFA, decreased KFA ROM and GCT, and increased approach velocity; and moderately associated with increased peak HFA (Table 8.10).

- Decreases in peak KAM were very largely associated with decreased peak KAA; largely associated with decreased peak KFM and IFPA; and moderately associated with decreased IC KAA and KIRM (Table 8.10).
- Decreases in peak KIRM were associated with increased peak HFA, and decreased peak KFM, decreased peak KAM, and lateral trunk flexion (Table 8.10).

8.5.3.2 CUT90 correlation values between changes in performance and injury risk variables with cutting mechanical variables

- Improvements in completion time were largely associated with increased peak KFM, FFC peak HBF, and decreased PFC GCT; and moderately associated with increased peak ADFA and ROM (Table 8.10).
- Improvements in FFC GCT were very largely associated with increased velocity at FFC, and decreased hip and knee flexion ROM; largely associated with increased exit velocity and lateral foot plant distance, and decreased peak KFA and peak HFA (Table 8.10).
- Improvements in exit velocity were very largely associated with increased velocity at FFC; largely associated with decreased FFC and PFC GCT, decreased HFA ROM, increased PFC mean HBF, peak VPF; and moderately associated with increased FFC peak KFM, and lateral foot plant distance (Table 8.10).
- Decreases in peak KAM were largely associated with decreased approach velocity and FFC peak HBF; and moderately associated with increased PFC mean and peak HBF, and decreased FFC peak KFM, and decreased velocity at FFC (Table 8.10).
- Decreases in peak KIRM were very largely associated with increased peak HFM; largely associated with decreased IC ADFA and pelvic rotation; and moderately associated with decreased peak and IC KAA, lateral foot plant distance and FFC peak VBF (Table 8.10).

Table 8.10. CUT45 and CUT90 associations between changes in performance and injury risk variables with changes in technical and mechanical variables

CUT45	Associated with	Correlation value	Descriptor
Improvements in completion time	1. Decreased GCT	1. $\rho = 0.691 \pm 0.306$, $p = 0.004$	1. Large
	2. Increased exit velocity	2. $r = -0.708 \pm 0.294$, $p = 0.003$	2. Very large
	3. Increased IC HFA	3. $\rho = -0.568 \pm 0.379$, $p = 0.022$	3. Large
	4. Decreased KFA ROM	4. $r = 0.469 \pm 0.424$, $p = 0.078$	4. Moderate
	5. Increased external HRA	5. $r = 0.510 \pm 0.407$, $p = 0.052$	5. Large
	6. Increased mean MLPF	6. $r = 0.400 \pm 0.449$, $p = 0.140$	6. Moderate
	7. Increased approach velocity	7. $r = -0.701 \pm 0.299$, $p = 0.004$	7. Very large
	8. Increased velocity at FFC	8. $r = -0.490 \pm 0.415$, $p = 0.063$	8. Moderate
Improvements in FFC GCT	1. Increased exit velocity	1. $\rho = -0.655 \pm 0.330$, $p = 0.008$	1. Large
	2. Decreased peak KFA	2. $\rho = 0.701 \pm 0.299$, $p = 0.004$	2. Very large
	3. Decreased KFA ROM	3. $\rho = 0.789 \pm 0.231$, $p < 0.001$	3. Very large
	4. Decreased peak ADFA	4. $\rho = 0.502 \pm 0.410$, $p = 0.056$	4. Large
	5. Increased approach velocity	5. $\rho = -0.644 \pm 0.336$, $p = 0.010$	5. Large
Improvements in exit velocity	1. Increased peak HFA	1. $\rho = 0.470 \pm 0.424$, $p = 0.077$	1. Moderate
	2. Increased IC HFA	2. $\rho = 0.545 \pm 0.391$, $p = 0.036$	2. Large
	3. Decreased KFA ROM	3. $r = -0.653 \pm 0.331$, $p = 0.008$	3. Large
	4. Decreased FFC GCT	4. $r = -0.686 \pm 0.309$, $p = 0.005$	4. Large
	5. Increased approach velocity	5. $r = 0.678 \pm 0.315$, $p = 0.005$	5. Large
	6. Increased velocity at FFC	6. $r = 0.772 \pm 0.245$, $p = 0.001$	6. Very large
Decreases in peak KAM	1. Decreased peak KFM	1. $\rho = 0.611 \pm 0.356$, $p = 0.016$	1. Large
	2. Decreased peak KAA	2. $\rho = -0.771 \pm 0.246$, $p = 0.001$	2. Very large
	3. Decreased IC KAA	3. $\rho = -0.479 \pm 0.420$, $p = 0.071$	3. Moderate
	4. Decreased KIRM	4. $\rho = -0.495 \pm 0.413$, $p = 0.061$	4. Moderate
	5. Decreased IFPA	5. $\rho = 0.600 \pm 0.362$, $p = 0.018$	5. Large
Decreases in peak KIRM	1. Decreased peak KFM	1. $r = -0.495 \pm 0.413$, $p = 0.061$	1. Moderate
	2. Increased peak HFA	2. $\rho = 0.466 \pm 0.425$, $p = 0.080$	2. Moderate
	3. Decreased peak KAM	3. $\rho = -0.495 \pm 0.413$, $p = 0.061$	3. Moderate
	4. Decreased IC lateral trunk flexion	4. $r = 0.485 \pm 0.418$, $p = 0.067$	4. Moderate
CUT90	**Associated with**	**Correlation value**	**Descriptor**
Improvements in completion time	1. Increased peak KFM	1. $r = -0.550 \pm 0.388$, $p = 0.034$	1. Large
	2. Increased peak HBF	2. $\rho = 0.557 \pm 0.385$, $p = 0.031$	2. Large
	3. Increased peak ADFA	3. $r = -0.459 \pm 0.428$, $p = 0.085$	3. Moderate
	4. Increased ADFA ROM	4. $r = -0.440 \pm 0.435$, $p = 0.101$	4. Moderate
	5. Decreased PFC GCT	5. $r = 0.669 \pm 0.321$, $p = 0.006$	5. Large
Improvements in FFC GCT	1. Increased exit velocity	1. $r = -0.633 \pm 0.325$, $p = 0.011$	1. Large
	2. Decreased peak HFA	2. $r = -0.525 \pm 0.400$, $p = 0.045$	2. Large
	3. Decreased HFA ROM	3. $r = 0.738 \pm 0.272$, $p = 0.002$	3. Very large
	4. Decreased peak KFA	4. $r = 0.677 \pm 0.315$, $p = 0.006$	4. Large
	5. Decreased KFA ROM	5. $r = 0.706 \pm 0.296$, $p = 0.003$	5. Very large
	6. Increased lateral foot plant distance	6. $r = 0.582 \pm 0.372$, $p = 0.023$	6. Large
	7. Increased velocity at FFC	7. $r = -0.846 \pm 0.179$, $p < 0.001$	7. Very large
Improvements in exit velocity	1. Decreased FFC GCT	1. $r = -0.633 \pm 0.343$, $p = 0.011$	1. Large
	2. Increased peak KFM	2. $r = 0.442 \pm 0.434$, $p = 0.099$	2. Moderate
	3. Decreased HFA ROM	3. $r = -0.569 \pm 0.379$, $p = 0.027$	3. Large
	4. Increased lateral foot plant distance	4. $r = -0.461 \pm 0.427$, $p = 0.084$	4. Moderate
	5. Increased PFC mean HBF	5. $r = 0.505 \pm 0.409$, $p = 0.055$	5. Large
	6. Increased peak VPF	6. $r = 0.722 \pm 0.284$, $p = 0.002$	6. Large
	7. Increased peak MLPF	7. $r = 0.496 \pm 0.413$, $p = 0.060$	7. Moderate
	8. Decreased PFC GCT	8. $r = -0.541 \pm 0.392$, $p = 0.037$	8. Large
	9. Increased velocity at FFC	9. $r = -0.713 \pm 0.291$, $p = 0.003$	9. Very large
Decreases in peak KAM	1. Decreased peak KFM	1. $r = 0.403 \pm 0.448$, $p = 0.011$	1. Moderate
	2. Increased PFC mean HBF	2. $r = -0.430 \pm 0.439$, $p = 0.110$	2. Moderate
	3. Decreased FFC peak HBF	3. $r = -0.664 \pm 0.339$, $p = 0.007$	3. Large
	4. Decreased approach velocity	4. $r = 0.550 \pm 0.388$, $p = 0.034$	4. Large
	5. Decreased velocity at FFC	5. $r = 0.496 \pm 0.413$, $p = 0.060$	5. Moderate
Decreases in peak KIRM	1. Increased peak HFM	1. $r = 0.753 \pm 0.261$, $p = 0.061$	1. Very large
	2. Decreased IC ADFA	2. $r = -0.522 \pm 0.401$, $p = 0.046$	2. Large
	3. Decreased peak KAA	3. $r = 0.446 \pm 0.433$, $p = 0.096$	3. Moderate
	4. Decreased IC KAA	4. $r = 0.495 \pm 0.413$, $p = 0.061$	4. Moderate
	5. Increased pelvic rotation	5. $r = 0.638 \pm 0.340$, $p = 0.010$	5. Large
	6. Decreased lateral foot plant distance	6. $r = 0.490 \pm 0.415$, $p = 0.064$	6. Moderate
	7. Decreased FFC peak VBF	7. $r = -0.468 \pm 0.424$, $p = 0.079$	7. Moderate

Key: GCT: Ground contact time; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; MLPF: Medio-lateral propulsive force; VBF: Vertical braking force; VPF: Vertical propulsive force; HFM: Hip flexion moment; KFM: Knee flexion moment; HFA: Hip flexion angle; KFA: Knee flexion angle; ADFA: Ankle dorsi-flexion angle; HRA: Hip rotation angle; ROM: Range of motion; IC: Initial contact; FFC: Final foot contact; IFPA: Initial foot progression angle

8.6 DISCUSSION

The aims of this study were two-fold: 1) to examine the biomechanical effects of a COD technique modification intervention on performance and multiplanar knee joint loads associated with increased ACL loading; and 2) to understand which technical and mechanical changes during cutting explained improvements in performance and reductions in knee joint loads. The primary findings were COD technique modification with externally directed verbal coaching cues resulted in statistically significant and meaningful improvements in cutting performance variables, which were statistically and meaningfully greater than the CG (Tables 8.2, 8.3, 8.6, 8.7, Figures 8.2 & 8.4), supporting the study hypotheses. The positive improvements in performance variables were primarily attributed to increases in velocity at key instances of the PFC and FFC, increases in propulsive force production in shorter GCTs, and decreased knee flexion ROM (Tables 8.4, 8.5, 8.8, 8.9, 8.10). In contrast to the study hypothesis, based on group means, no statistically significant and meaningful changes in multiplanar knee joint loads were observed post-intervention for the IG (Tables 8.2, 8.3, 8.6, 8.7, Figures 8.3 & 8.5). However, closer inspection amongst the individual data showed considerable individual variation (i.e., positive and negative responders) and mixed responses to the intervention (Figures 8.3 & 8.5). Generally, participants who displayed initially (pre-intervention) high multiplanar knee joint loads and thus considered potentially "high-risk", responded positively and demonstrated reductions in knee joint loads (Figures 8.3 & 8.5). Consequently, COD technique modification appears to be a training strategy effective for improving COD performance, while potentially reducing knee joint loads in "higher-risk" athletes.

8.6.1 Effect on performance variables

A plethora of researchers have shown that COD speed training improves completion time (77, 81, 87, 332, 347, 348, 460, 630); however, the technical and mechanical mechanisms explaining the improvements in COD performance are relatively unknown. This study is one of the first to provide a novel insight into the mechanical and technical changes underpinning improvements in 45° and 90° cutting performance. Table 8.10 provides the associations between technical and mechanical changes with performance variables, and the pre-to-post changes in cutting technical and mechanical variables are presented in Tables 8.5 and 8.9. The key and primary changes in cutting mechanics were as follows: increased velocity at key instances of the PFC and FFC, increased mean and peak propulsive forces in shorter FFC GCTs, and reduced knee flexion ROM which were moderately to very largely associated with improvements in cutting performance (Table 8.10). The fact that the abovementioned variables were associated with improvements in COD performance are unsurprising because they have been identified as key determinants of faster COD performance (115, 143, 213, 228, 275, 346, 530, 531, 597), and key variables associated with faster performance in Chapter 5.

For example, Hader et al. (213) indicated greater minimum velocities and velocity maintenance were strong determinants of faster 45° and 90° cut performance, thus increasing an athlete's ability to cover distance in shorter times. Greater propulsive forces increase impulse which, based on Newton's 2nd law of motion and the impulse-momentum relationship, leads to greater changes in momentum, thus velocity (56, 634), and this was confirmed by the moderate association between MLPF changes and exit velocity (Table 8.10). Additionally, shorter GCTs are associated with faster performance (143, 338, 346, 498, 531, 596, 597), which is advantageous because the athletes will spend less time braking and propelling, thus less time performing the COD action, and potentially greater utilisation of the SSC (55, 596). IG participants demonstrated smaller ADFA ROMs for CUT45 performance post-intervention (Table 8.5). Reduced ankle dorsi-flexion has been reported to play a critical role in early acceleration sprint performance (45, 46, 83), mostly likely contributing to a stiffer ankle joint; thus, potentially contributing to the faster performance observed in the present study. Finally, reduced KFA ROM is key for reducing GCT and increasing exit velocity, as confirmed by the results from this study, Chapter 5, and Dai et al. (115). This potentially results in a rapid transition from braking (eccentric) to propulsion (concentric) and therefore facilitating a shorter GCT and greater utilisation of the SSC (increased elastic energy) to permit more effective force generation (55, 317).

High propulsive forces (143, 228, 530, 531, 596) over short GCTs (143, 338, 346, 498, 531, 596, 597) have been identified as strong determinants of faster COD performance. The finding of increased propulsive force production in shorter GCTs following COD technique modification training is a positive outcome (Table 8.5 & 8.9), and is similar to de Hoyo et al. (121) who found 10-weeks eccentric overload training resulted in likely to almost certain increases (ES = 0.48-1.24) in relative peak braking and propulsive forces, shorter braking and propulsive times thus shorter GCTs, and increases in relative total and braking impulse during side-step cutting. Unfortunately, de Hoyo et al. (121) did not examine completion time or exit velocity; however, improvements in completion time training have been observed following flywheel eccentric training (199, 560); thus, it is speculated that the improved side-step kinetics from flywheel training should result in improved COD performance. Similarly, corroborating the mechanism of improved completion time in the present study, Bencke et al. (33) reported improvements in VPF (ES = 0.41), reductions in GCT (ES = 0.94), and reductions in concentric phase duration (ES = 0.94) following 12-weeks neuromuscular warm-up training (jump-landing, unilateral squats, hamstring pulls, hip abductions, and one leg hopping).

Additionally, to the best of the author's knowledge, King et al. (292) is the only other research group to document changes in cutting biomechanics and some performance indices following a training intervention. No significant changes in approach velocity (p = 0.434, ES = 0.07) were observed following a multicomponent training programme (resistance training, running drills, and COD drills with focus on intersegmental control); however, a reduction in GCT was observed (ES = 0.30) which is a key determinant of COD performance. Unfortunately, King et al. (292) did not examine completion time or exit velocity, but did report reductions in lateral trunk flexion, increased pelvic rotation towards the intended direction, increased COM to COP distance, increased ankle plantar-flexor moments, and decreased peak KFA, which have all been associated with faster cutting completion times (115, 228, 346, 597). Although no significant changes in lateral trunk flexion or foot plant distance were observed in the present study (Tables 8.5 & 8.9), similar to King et al. (292), participants post-intervention in the IG demonstrated increased pelvic rotation, decreased peak KFA, and increased ADFMs, which have previously been associated with faster cutting performance (115, 228, 346, 597). Nevertheless, the results from this study confirm that COD speed and technique modification training with external verbal coaching cues regarding technique result in positive improvements in side-stepping performance. Therefore, practitioners should consider implementing this form of training in their training programmes when working with multidirectional athletes.

8.6.2 Effect on injury risk variables

A key strategy to reduce potential non-contact ACL injury risk is to reduce multiplanar knee loads which have the potential to strain the ACL (126, 183, 241, 246, 327, 328, 435, 535). COD technique modification is one training strategy that has been shown to acutely reduce knee joint loads during cutting (76, 115, 127), while reductions in peak KAMs have also been observed following 6-weeks technique modification during COD (126, 277). In the present study, no statistically significant interaction effects were observed for any knee joint loads (Tables 8.2 & 8.6), and pre-to-post changes in multiplanar knee joint loads for the IG were non-significant with trivial to small effect sizes (Tables 8.3 & 8.7, Figures 8.3 & 8.5). These results contrast to Dempsey et al. (126) and Jones et al (277); however, it is important to note that the IG increased their velocity at key instances of the PFC and FFC which can directly influence and amplify knee joint loads (148, 290, 403, 574). Secondly, and importantly, these two aforementioned studies did not contain a CG. The present study contained a CG which notably demonstrated a significant, large increase in peak CUT45 KAMs, and a non-significant yet small increase in CUT90 peak KAMs (Tables 8.3 & 8.7, Figures 8.3 & 8.5). Although it is difficult to fully explain this finding, Staynor et al. (535) also reported greater KAMs and KIRMs for a CG post-intervention (ES = 0.36-0.56), which was potentially attributed to the lack of specific injury mitigation training performed in-season by the CG, while the IG remained unchanged. Thus, the lack of specific COD training with corrective feedback and greater KAA observed for the CG may partially explain why the CG increased their peak KAMs post-intervention in the present study.

Dempsey et al. (126) is the only other study to investigate the effects of technique modification training on side-step knee joint loads and found 6-weeks training produced a 36% significant reduction in peak KAMs ($p = 0.034$, ES = 0.58-0.78) which was attributed to positive changes in lateral trunk flexion ($r = -0.377$, $p = 0.135$) and lateral foot plant distance ($r = -0.468$, $p = 0.025$). It is worth noting, however, that peak KIRMs remained unchanged. The findings by Dempsey et al. (126) are in contrast to the present study that observed no significant and meaningful reductions in multiplanar knee joint loads for the IG following technique modification based on group means. However, this discrepancy could be attributed to differences in the delivery of the training intervention and methodological differences. For example, Dempsey et al. (126) had considerably lower athlete to coach ratios of 1:1 and 2:1 and also used time warp video feedback between repetitions to provide biofeedback regarding the technical changes. Conversely, this was not feasible for the present study whereby higher athlete to coach ratios were present (typically 5:1) and no video feedback was given regarding technique. This lack of biofeedback may partially explain why there were no meaningful reductions in knee joint loads (based on group means) in the present study for the IG.

Crucially, an integral difference between the two studies were the targeted technical modifications, with the goal of Dempsey et al. (126) to reduce multiplanar knee joint loads, in contrast to the present study which also simultaneously aimed for performance improvements. Dempsey et al. (126) focused primarily on adopting an upright trunk posture in the frontal plane and reducing lateral foot plant distance with the use of line markings for acceptable foot placement. While the present study did attempt to alter frontal plane trunk control, participants were instructed to "cushion and push the ground way", with no restrictions on lateral foot plant distance because of the potential detrimental effects narrowing lateral foot plant distance may have on MLGRF propulsion and subsequent performance (228, 261, 272, 597). Conversely, the present study attempted to increase initial KFAs, improve frontal plane knee control, and encourage PFC dominant braking strategies (for CUT90). Finally, it is worth noting that Dempsey et al. (126) performed the side-steps at a controlled approach velocity, and did not examine completion time, GCTs, or exit velocity. Conversely, the present study considered the performance implications in order to improve athlete and coach adherence, and the athletes crucially increased their approach and FFC velocity.

Based on group means, no statistically significant and meaningful changes in multiplanar knee joint loads were observed post-intervention for the IG (Table 8.3 & 8.7). In applied and clinical settings, however, practitioners do not work with group means but individuals. Figures 8.3 & 8.5 illustrate the IG individual responses in multiplanar knee joint loads following the training intervention which shows considerable individual variation (i.e., positive and negative responders) and mixed responses to the intervention. This finding is similar to previous research that has shown individual variation in response to injury mitigation training interventions (110, 122, 186, 392, 457). Generally, participants with initially high multiplanar knee joint loads and thus considered to be potentially at higher and greater injury risk (241, 242), responded positively and demonstrated reductions in knee joint loads (Figures 8.3 & 8.5). This observation is similar to previous research that found "higher-risk" female athletes responded favourably to injury mitigation training by displaying greater reductions in landing KAMs compared to "lower-risk" athletes (122, 186, 392). The present study, however, is the first to have examined the individual changes in knee joint loads during cutting following a COD technique modification training intervention, highlighting that an individual approach is needed because inferences based on group means only may conceal potentially meaningful information (110, 186).

Specifically, more participants demonstrated positive reductions in multiplanar knee joint loads during CUT90 compared to CUT45 (8 vs. 5) which were greater than the SWC and SEM (Figures 8.3 & 8.5). Although some participants did increase their peak KAMs and KIRMs following the training intervention (Figures 8.3 & 8.5), it is important to note that some of the participants had low knee joint loads initially and the knee joint loads they displayed post-intervention may not be considered as potentially "high-risk" and hazardous (122). It is important to note that some participants in the IG displayed knee adduction moments during some of the CUT45 trials pre-intervention which may explain some of the initial low KAMs for some participants. This observation substantiates Sigward and Powers (515) who also found some male athletes displayed knee adductor (varus) moments during a CUT45 task. Furthermore, it should also be acknowledged that the majority of athletes increased their approach and exit velocities, which can also amplify knee joint loads (148, 290, 403, 574). Consequently, COD technique modification appears to be a training strategy effective in improving COD performance, while potentially reducing knee joint loads in "higher-risk" athletes.

In addition to multiplanar knee joint loads, changes in postures and mechanics associated with increased knee joint loads were also assessed in the present study. In contrast to Dempsey et al. (126), COD technique modification training in the present study did not result in significant or meaningful changes in lateral foot plant distance or lateral trunk flexion (Tables 8.5 & 8.9). The finding that lateral foot plant distance did not change, based on group means, is unsurprising because this was not a specific targeted technical change. Conversely, it is somewhat surprising that lateral trunk flexion angles were not significantly and meaningfully different because athletes were specifically given the verbal cue to "lean and face towards the intended direction of travel". For example, Staynor et al. (535) observed reductions in lateral trunk flexion angles following mixed training which combined BW plyometric, resistance, and balance exercises, while King et al. (292) found a three-phase programme which focused on intersegmental control and strength, intersegmental control during running and COD reduced lateral trunk flexion angles during cutting (ES = 0.79). Potentially, external verbal cues do not provide a sufficient stimulus to evoke frontal plane trunk control changes and thus, increases in physical capacity and intersegmental control is needed through direct conditioning (292, 535, 602). However, closer inspection amongst the individual data show (Figure 8.6) positive and negative responders for changes in lateral trunk flexion angle, with eight and seven athletes positively reducing their lateral trunk flexion angles greater than the SWC for CUT45 and CUT90, respectively. As such, the mixed variation in response to the training intervention conceals potentially meaningful differences based on group mean analysis, and highlights that an individual approach is needed when monitoring changes in COD biomechanics (110, 186).

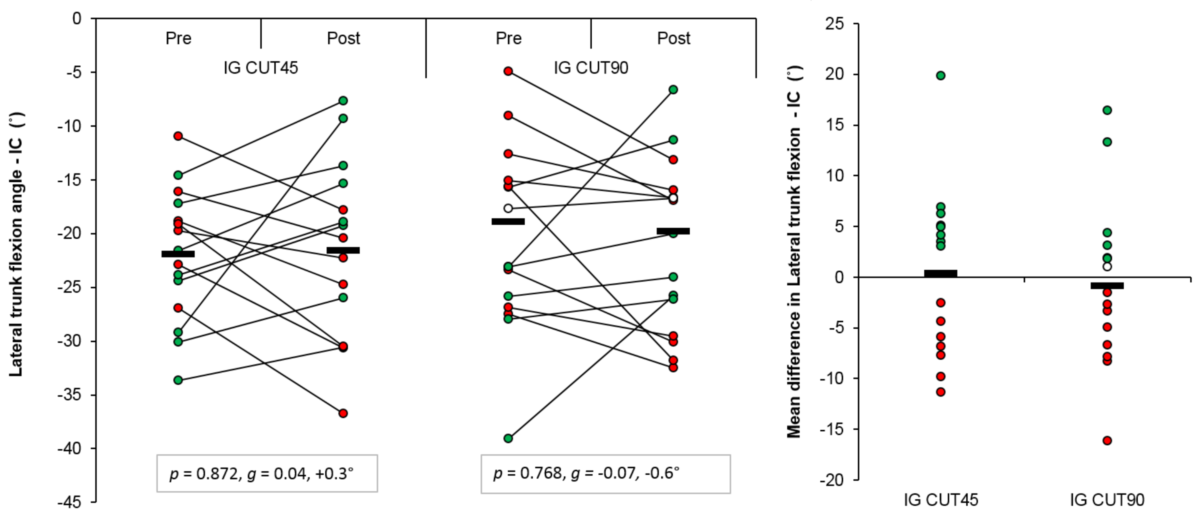


Figure 8.6. Pre-to-post changes in CUT45 and CUT90 lateral trunk flexion angles for IG: Intervention group; IC: Initial contact; green = positive responder; red = negative responder; white = non-responder.

Cutting postures with limited knee flexion and high impact GRFs are identified as "high-risk" characteristics of non-contact ACL injury (52, 63, 94, 204, 270, 291, 298, 310, 376, 426, 583) and associated with increased knee joint loads (42, 43, 343, 615). Although no meaningful reduction in peak VBF were observed, a positive outcome following the intervention was a small increase in initial KFA at FFC (Tables 8.5 & 8.9), though this change was ~2-3˚. This technical change may be attributed to the coaching cues to "cushion over WA" and "push the ground away". Critically, however, increased IC and peak KAA was observed following the intervention (small to large ES), based on group means (Tables 8.5 & 8.9). This increase could potentially be attributed to the increased approach velocity, whereas Sigward and Powers (519) suggest that an internally rotated lower-extremity position might be adopted by athletes to encourage the COM of the body further away from the COP, and to facilitate the directional change to the intended direction of travel through a combination of rotations of the lower-limb joints. However, the results of Chapter 5 and previous research (228, 346) show no meaningful relationships between KAA and faster cutting performance. Nevertheless, these findings highlight the difficulty in improving frontal plane control during cutting using externally directed verbal coaching cues only. Potentially, athletes would benefit from external hip rotator strengthening to improve frontal plane knee control during side-stepping (287, 292, 341, 432, 535).

A unique aspect of the present study was to understand the technical and mechanical changes in cutting technique associated with changes in knee joint loads (Table 8.10). The results from this study provide insight into which potential side-step cutting deficits increase and decrease knee joint loads, and thus could be used to inform future directions of training. Specifically, decreases in peak and IC KAA were moderate to largely associated with decreases in CUT45 and CUT90 peak KAMs and KIRMs, and are therefore specific deficits to target when aiming to reduce multiplanar side-step knee joint loads. In addition, a noteworthy observation was increased IC KFA was moderately associated with reductions in CUT45 KAM, decreased lateral trunk flexion was also moderately associated with decreases in CUT45 KIRM, and decreases in FFC VBF were moderately associated with reductions in CUT45 KAM and CUT90 KIRM. These aforementioned variables appear to be specific deficits to target in future training interventions. Finally, decreases in approach velocity and FFC velocity were largely and moderately associated with reductions in CUT90 KAM, respectively, but are not recommended deficits to target due to their importance for performance (148)(Chapter 5).

8.6 LIMITATIONS

It is worth noting that the present study has several limitations. Firstly, it should be acknowledged that only a 45° and 90° side-stepping COD task was investigated with a right limb push-off. As the biomechanical demands are angle- and task-dependent (56, 148, 411), caution is advised extrapolating the findings from the present study to CODs of different angles and actions. As such, further research is necessary that investigates the effect of COD technique modification on sharper CODs and different COD actions (XOC, split step). Due to laboratory configuration, the present study only examined side-stepping on the athletes' right limb, which in most cases was their preferred kicking and push-off limb. Athletes can display performance deficits and subtle differences in COD mechanics between limbs (145). Thus, it was unfortunate that the present study did not examine changes in the left limb which was predominantly the non-dominant limb for the sample, where arguably this method of training may have had a greater effect. Future research should consider the effects of COD technique modification on both dominant and non-dominant limbs. Similarly, it is worth acknowledging that the cutting actions were examined in a pre-planned, controlled laboratory environment. Although outside the scope of the present study, it is unknown how this method of training effects on-field performance (i.e., tackle break success, duel success, number of evasions etc.) and unanticipated cutting mechanics, and is a recommended avenue for further research.

As stated in the discussion, a key methodological consideration which may explain the contrasting findings between the present study and Dempsey et al. (126) is the absence of biofeedback and the differences in coach to athlete ratio regarding technique. Unfortunately, in the present study, due to logistical constraints, it was not feasible to fully provide individualised feedback with video technology within 30 minutes, for group sizes typically of five or greater. Additionally, due to time constraints, there was no initial pre-screening of individuals to specifically identify targeted deficits to inform technique modification training, and the individuals' knee joint loads pre-intervention were unknown, so it was difficult to specifically target "high-risk" athletes. This absence was due to a two-week time constraint to conduct the pre-intervention 3D motion analysis. Potentially the CMAS could have been used to conduct some quicker and initial screening of athletes to inform technique modification training for the intervention. Furthermore, although outside the scope of the study, changes in muscle activation were unexamined. Wilderman et al. (606) has found an increase in hamstring muscle activation following COD speed and footwork training. Though changes in multiplanar knee joint loads were non-significant and trivial to small in the present study (Tables 8.3 & 8.7, Figures 8.3 & 8.5), speculatively there may have been increased muscle activation of the hamstrings, gluteal muscles, and soleus, which may have the potential to help unload the knee ligaments (126, 138, 328, 341). Future research should monitor muscle activation changes in addition to the biomechanical effects of COD technique modification.

Finally, the aim of this study was to solely examine the biomechanical effects of COD technique modification on performance and injury risk perspectives. In applied settings, however, athletes would perform a mixed and multicomponent training programme (148, 432). For example, in a recent position statement on ACL injury risk mitigation training, it was recommended that a mixed multicomponent programme which incorporates strength, balance, trunk control, plyometrics, and COD/agility training is performed (432), and these modalities would commonly feature in an athlete's holistic strength and conditioning programme. Balance training has been shown to be effective in reducing side-stepping knee joint loads (92, 93, 425), potentially attributed to favourable changes in muscle activation patterns of the hamstring, trunk, and hip musculature (92, 425). Additionally, reductions in side-stepping knee joint loads have been demonstrated as a result of dynamic trunk stability training (602). Consequently, the present study may have benefited from the addition of supplemental conditioning (balance and dynamic trunk stability exercises) to help improve muscular activation and postural control. As such, future research to determine the effects of a mixed multicomponent training on COD biomechanics is needed to increase the ecological validity to the field and "real-world" environments. Lastly, it is unknown whether the technique can be maintained for extensive periods and what happens to cutting biomechanics when you discontinue this method of training. Thus, further investigation into short- and long-term retention of modified cutting technique is required, to improve our understanding of training prescription and dosages.

8.7 CONCLUSION

In conclusion, this is the first study to comprehensively examine the biomechanical effects of COD technique modification with externally directed verbal coaching cues on performance and injury risk, finding statistically significant and meaningful improvements in side-step cutting performance variables. The positive improvements in performance were primarily attributed to increases in velocity at key instances of the COD, increases in propulsive force production over shorter GCTs, and decreased KFA ROM. Based on group means, COD technique modification was ineffective regarding potential injury risk, resulting in no statistically significant and meaningful changes in multiplanar knee joint loads post-intervention for the IG. However, on an individual basis, considerable variation was observed in response to the intervention (i.e., positive and negative responders) and mixed responses to the intervention. Generally, athletes who displayed initially high multiplanar knee joint loads (pre-intervention) and thus considered potentially "high-risk", responded positively and demonstrated reductions in knee joint loads; highlighting the importance of an individual approach when monitoring the effectiveness of training interventions. Conversely, athletes with initially low multiplanar knee joint loads tended to increase their multiplanar knee joint loads, but the joint loads they displayed post-intervention may not be considered as potentially hazardous and "high-risk". This highlights the need to pre-screen athletes prior to technique modification training to enable individual technique deficits to be targeted during the intervention. Consequently, COD technique modification appears to be a training strategy effective for improving COD performance, while potentially reducing knee joint loads in "higher-risk" athletes. Practitioners can therefore consider incorporating this form of training into their programmes to enhance side-step cutting performance and potentially reduce knee joint loads in "higher-risk" athletes.

CHAPTER 9: SYNTHESIS OF FINDINGS

9.1 Global discussion

The primary aims of the thesis were three-fold: 1) to identify the biomechanical determinants of performance and injury risk during cutting, in order to better understand the potential "performance-injury conflict"; 2) to validate a qualitative screening tool against 3D motion analysis for evaluating cutting movement quality and identifying athletes who display high peak KAMs; and 3) to understand the biomechanical effects of a COD technique modification intervention, with the aim of reducing knee joint loads while maintaining or enhancing cutting performance.

In order to investigate these aims, however, it was important to establish appropriate data collection and analysis procedures, while also determining between-session reliability measures. Initially, two pilot studies were conducted (Appendix 2.1 & 2.2) regarding the effect of low-pass filtering COF combinations on joint moments and averaging methods of obtaining discrete data. Substantiating previous research (44, 304, 351, 493, 558), low-pass filtering COF combination significantly affected KAM and HFM magnitudes, while KAM ranking was also affected. Interestingly, COF combination had a negligible effect on KFMs and ADFMs. Based on the results of the pilot study and recommendations from Roewer et al. (493), it was concluded that a 15-25 Hz was the most suitable COF combination going forward because it demonstrated good agreements in terms of KAM participant ranking against matched COF combinations, and it preserved the signal to a greater extent; thus, preserving potentially meaningful physiological information. Additionally, it was concluded that discrete data for cutting outcome values should be obtained based on the average of individual trial peaks because it is not influenced by misalignments and variations in trial peak locations, in contrast to the peak from average profile. As such, these findings informed the data analysis procedures for the subsequent studies.

Trial size is an important methodological consideration in the design of human movement experiments (200, 220, 264, 370, 374, 383, 553) and establishing between-session reliability is fundamental for establishing "real" changes in performance (5). Reliability measures for COD biomechanics are rarely reported across the literature (Table 4.1), and only a limited number of studies have comprehensively examined the between-session reliability (5, 370, 497), while only one study has considered the effect of trial size (370). As such, Chapter 4 aimed to answer the following research question:

What is the effect of trial size on the within- and between-session reliability measures and outcome values for cutting biomechanics?

The results from Chapter 4 confirmed that cutting joint angle, GRF (excluding FFC GCT), and KFM variables demonstrate high reliability, while KIRMs, KAMs, and KAAs demonstrate high variability and bias between sessions. Thus, caution is recommended when monitoring changes in KIRMs, KAAs, and KAMs. Additionally, there appeared to be no additional benefit of increasing trial sizes, with minimal differences in reliability measures and outcome values between trial sizes. As such, trial sizes of three to five are adequate to capture reliable cutting biomechanical data, with no apparent requirement for greater trial sizes. Using lower trial sizes may reduce the time and financial costs associated with 3D motion analysis testing in clinical settings and reduce athlete fatigue during maximal effort testing.

Despite the importance of directional changes for sports performance and its association with ACL injury risk, it is somewhat surprising that the majority of studies into COD biomechanics investigate performance (143, 275, 338, 346, 498, 530, 531, 596, 597) and injury risk determinants (117, 127, 139, 267, 272-274, 301, 358, 516, 519, 625) independently. Prior to the thesis, there was preliminary evidence from one study (228), which indicated that techniques and mechanics associated with faster cutting performance are in direct conflict with reduced KAMs, while Fox (183) in a recently published review also discussed the concept of a conflict between performance and ACL injury mitigation. Athletes are less likely to adopt safer cutting strategies if they negatively affect performance (183, 228). Therefore, this conflict is problematic for practitioners who aim to improve their athletes' COD performance while minimising injury risk. Consequently, a primary aim of the thesis (Chapter 5) was to expand on the work of Havens and Sigward (228) and better understand the potential "performance-injury conflict" by answering the following research question:

Do COD techniques and mechanics which result in faster performance have a concurrent increased risk of injury?

Additionally, because emerging research has highlighted that the PFC plays a major role in deceleration (147) prior to push-off, and is associated with faster COD performance (143, 202, 275) and potentially lower knee joint loads (272-274), a further research question was:

Do braking strategies which emphasise PFC braking forces help alleviate knee joint loading and facilitate faster performance during cutting?

The primary findings from Chapter 5 were that techniques and mechanics associated with faster cutting performance (i.e., faster velocities over key instances of the PFC and FFC, greater FFC braking forces over short GCTs, greater KFMs, smaller hip and KFAs and ROM, wider lateral foot plants, and greater IFPAs) are in direct conflict with safer cutting mechanics (i.e., reduced knee joint loading), and support the concept of a “performance-injury conflict”. Therefore, from a performance perspective, practitioners may consider developing their athletes’ speed and force production capacity, while coaching and cueing cutting techniques such as the adoption of “stiffer” hip and knee strategies, wider lateral foot plants (to “push the ground away”) and to rotate their whole-body COM towards the intended direction of travel. However, practitioners must be mindful that while these desired techniques and mechanics are optimal for faster cutting, there may be a concurrent increase in multiplanar knee joint loads and potential non-contact ACL injury risk. Thus, it is imperative that practitioners are conscious of this conflict when instructing cutting techniques to optimise performance while minimising knee joint loading, and ensure that athletes have the physical capacity to tolerate and support knee joint loading (i.e., neuromuscular control, co-contraction, and rapid force production) (39, 138, 272, 301, 328, 341, 410, 432, 544, 591).

In addition to physical preparation, it is worth noting that some key deficits may help alleviate, but not solve, the “performance-injury conflict”. For example, KAA was linked with greater knee joint loads but offered no associated performance benefits, while PFC braking dominant strategies (i.e., greater PFC HBF/ratios) and minimising lateral trunk flexion were factors associated with faster performance and lower knee joint loads (Chapter 5). Therefore, a key application from this study is practitioners should consider coaching PFC dominant braking strategies and implementing training strategies that reduce KAA and lateral trunk flexion angles, to help reduce knee joint loading and potentially facilitate effective cutting performance.

Based on the literature reviews in Chapter 2.3.3 and 2.3.4, cutting performance and injury risk deterministic models were created. In light of the findings observed in Chapter 5, the deterministic models have been updated. Consequently, Figures 9.1 and 9.2 illustrate the updated changes, with green text indicating which results from Chapter 5 substantiate the literature, and blue text indicating novel findings. Based on these deterministic models, a side-stepping technical model (Figure 9.3) has been created which practitioners can use for enhancing performance and reducing knee joint loads and thus, may assist in the development of more effective COD speed and ACL injury mitigation programmes. However, it should be noted that the side-step technical models presented in Figure 9.3 is for mitigating ACL injury risk; thus, optimal technical models for mitigating ankle and groin injuries are unclear and warrant future inspection.

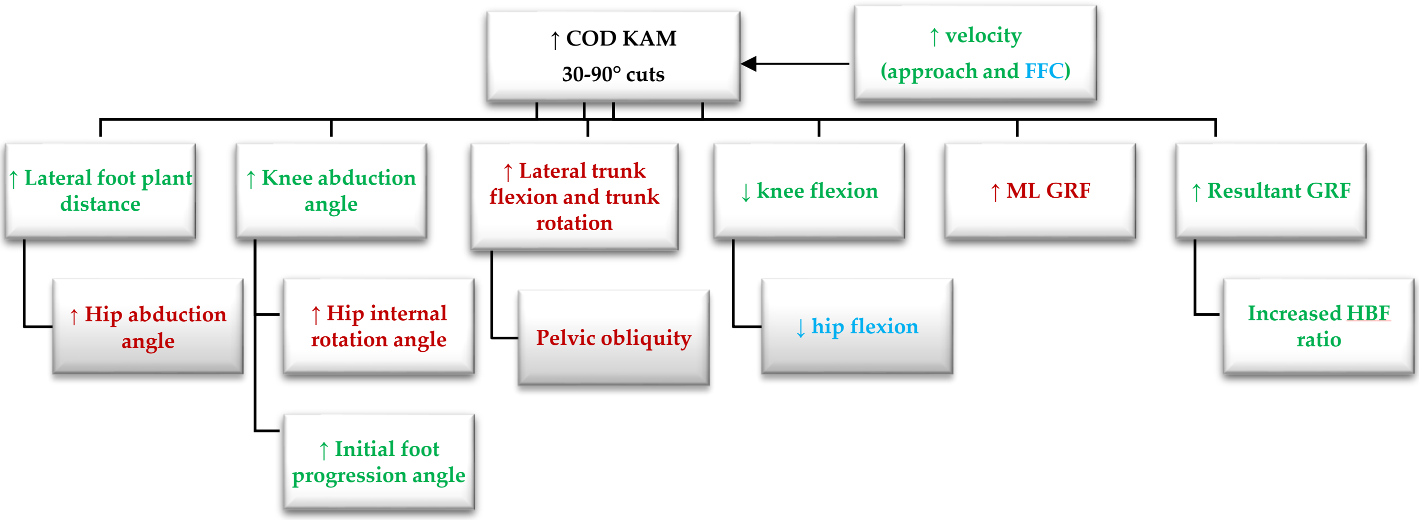


Figure 9.1. Updated KAM deterministic model during side-step cutting. Green illustrates which results from Chapter 5 substantiate literature, while blue indicates novel findings (Dempsey et al. (126, 127); Havens and Sigward (228); Jones et al. (272); Kristianslund et al. (301); Sigward and Powers (519); McClean et al. (358); Jamison et al. (267); Frank et al. (189); Donnelly et al. (141); David et al. (117); Staynor et al. (534); Weir et al. (593))

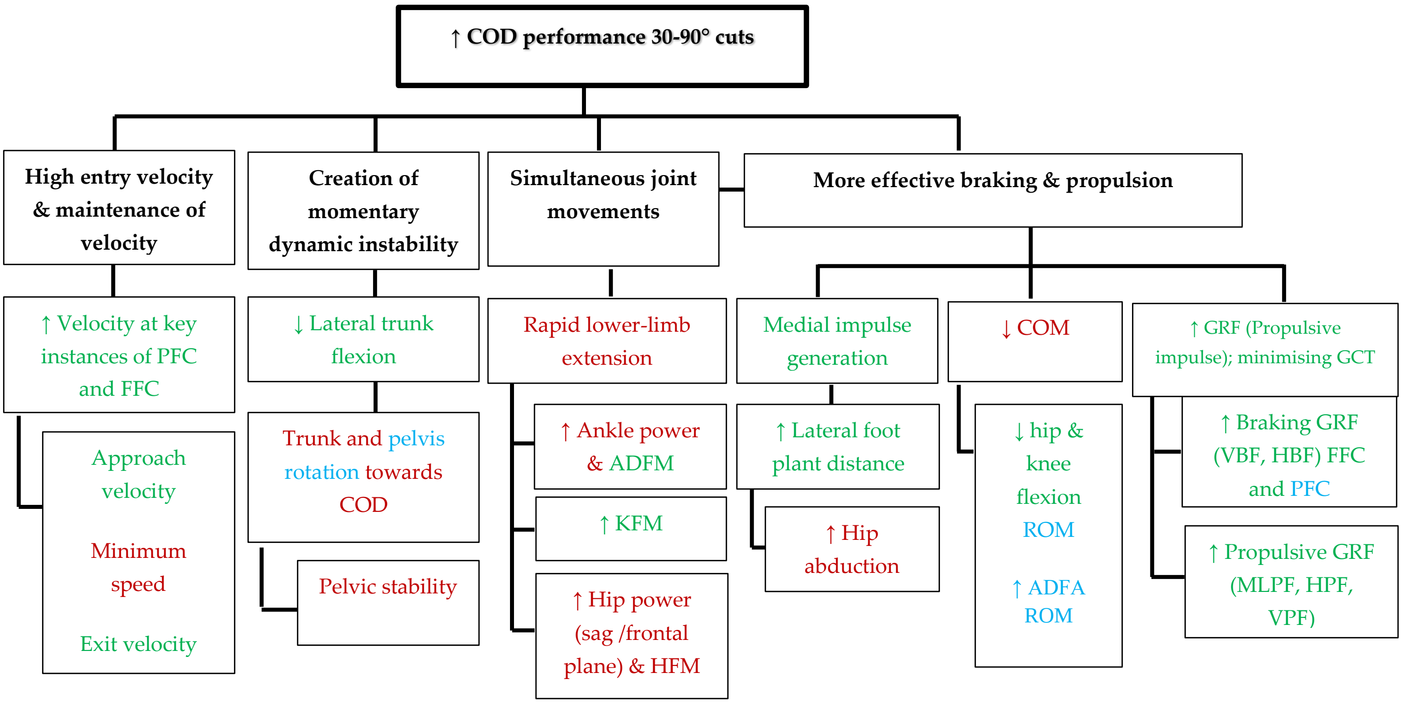


Figure 9.2. Updated Cutting performance deterministic model during side-step cutting. Green illustrates which results from Chapter 5 substantiate literature, while blue indicates novel findings (Hader et al. (213); Inaba et al. (261); Jones et al. (272, 275); Havens and Sigward (228); Andrews et al. (11); Spiteri et al. (530, 531); Dos'Santos et al. (143, 146, 147); Graham-Smith et al. (202); Maloney et al. (338); Marshall et al. (346); Sasaki et al. (498); Welch et al. (596, 597))

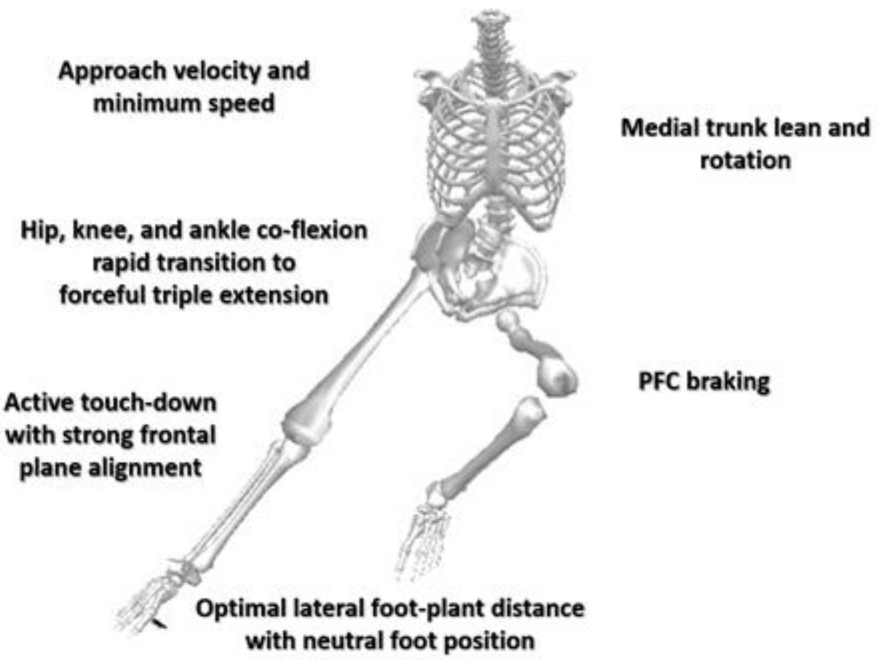


Figure 9.3. Side-step cutting technical model for faster and safer performance

Following on from Chapter 5 and the updated deterministic models, a further aim of the thesis was to validate a cutting qualitative screening tool against 3D motion analysis. The specific items of the screening tool were based on determinants of greater peak KAMs (Figure 9.1). Although 3D motion analysis is the gold standard for evaluating movement, it is a time-consuming and expensive process, with limited applicability in the field (187, 241). Consequently, practitioners require a qualitative screening tool for evaluating cutting mechanics to identify athletes who display "high-risk" and aberrant mechanics, so individualised training programmes can be created. At the time of writing the thesis, qualitative screening tools existed for jump-landing assessments and single-leg squatting, but no such tool existed for cutting. This was a key absence especially for practitioners who work in cutting dominant sports, such as rugby, handball, American football, and soccer. As such, Chapter 6 aimed to answer the following research question:

Can a qualitative screening tool identify athletes with "high-risk" movement mechanics and high peak KAMs during cutting?

The primary aim of Chapter 6 was to expand on the preliminary investigation which initially validated the CMAS (278), by investigating a larger sample size, and using an additional camera at a greater sampling rate (Chapter 6). Substantiating the results of the earlier investigation, a strong relationship was observed between CMAS and peak KAMs, and trials and participants who displayed greater CMASs typically displayed "higher-risk" postures and greater multiplanar knee joint loads (Chapter 6). The results confirm that the CMAS is a valid and reliable screening tool to identify athletes who display "high-risk" movement mechanics and knee joint loads during side-step cutting, with the use of only three high-speed cameras and free video analysis software. Consequently, the CMAS offers practitioners a cost-effective, easily applicable method to evaluate cutting mechanics in the field, which has been validated against the gold standard of 3D motion analysis. Practitioners can therefore directly identify specific deficits and "high-risk" athletes to create individualised training programmes and monitor the effectiveness of such programmes by revaluating CMAS performance.

Finally, a significant strength of the thesis was that two COD speed and technique modification interventions were conducted. Practitioners are interested in training interventions that can improve COD performance and mitigate injury risk. Preliminary evidence indicated that COD technique modification training is an effective strategy in reducing knee joint loads but the implications on performance are not well understood and no study has contained a CG (76, 115, 126, 127, 277). Thus, Chapter 8 aimed to answer the following research question:

What are the biomechanical effects of a six-week COD technique modification intervention on performance and injury risk?

Initially, a six-week intervention study (Chapter 7) was performed in an applied field-setting with professional male youth soccer players, with the validated CMAS (Chapter 6) used to monitor changes in cutting movement quality. The primary findings were that 6-weeks COD speed and technique modification training with externally directed verbal coaching cues improved cutting performance and movement quality in this population. Practitioners working in soccer could easily integrate this form of training into the warm-ups of field-based sessions (2 × 20-minute sessions per week) with minimal equipment, and importantly can use the CMAS to monitor such changes in movement quality, demonstrating its applicability in the field.

Chapter 7 served as a feasibility study before completing a similar COD technique modification intervention (Chapter 8) but comprehensively monitoring the biomechanical changes with 3D motion and GRF analysis. Chapter 8 is the first study, to the best of the author's knowledge, to examine the biomechanical effects of cutting technique modification on both performance and knee joint loads, while also containing a CG. The primary findings were that COD technique modification training with externally directed verbal coaching cues resulted in significant and meaningful improvements in side-step cutting performance. When examining group means, no statistically significant or meaningful changes in multiplanar knee joint loads were observed; however, there was considerable individual variation in response to the training intervention. It would, however, be erroneous to assume that all athletes would respond favourably and to the same magnitude following the training intervention (110, 186). The programme appeared effective in reducing multiplanar knee joint loads in athletes considered "higher-risk", substantiating the results of previous research that demonstrated greater reductions in landing KAMs for "higher-risk" female athletes following injury mitigation training (122, 186, 392). Conversely, in the present study, increased knee joint loads in athletes with initially low multiplanar knee joint loads in response to the programme were observed, though it must be stressed that the post-intervention knee joint loads were not considered "high" and potentially hazardous. Consequently, the results from this study highlight the importance of an individual approach when monitoring changes in biomechanics following a training intervention, because inferences based on group means may conceal potentially meaningful information (110, 186).

Dempsey et. (126) reported reductions in cutting peak KAMs as a result of technique modification and video feedback. Unfortunately, it was not feasible to fully provide individualised feedback with video technology (within 30 minutes), for group sizes typically of 5 or greater (Chapter 8). Additionally, due to time constraints, there was no initial pre-screening of individuals to specifically identify targeted deficits to inform future training (Chapter 8). This absence was due to the two-weeks' time constraints to conduct the initial pre-intervention 3D motion analysis testing, which may potentially explain the mixed responses as a result of the intervention. For some athletes, video feedback may have been useful to reinforce and correct cutting techniques in addition to the external verbal coaching cues. Potentially the CMAS could have been used to conduct initial pre-screening of athletes to inform technique modification training for the intervention (Chapter 6). Consequently, future research should attempt to replicate the technique modification study with the use of biofeedback and initial pre-screening of athletes to inform the technique modification training.

The aim of the intervention studies was to solely examine the effects of COD technique modification on performance and injury risk (knee joint loads). In applied settings, however, athletes would perform a mixed multicomponent strength and conditioning programme (148, 432), to ensure that athletes not only have the physical capacity to perform COD, but the capacity to perform a range of other movements in sport (jumping, sprinting, deceleration, kicking, throwing etc). In a recent position statement for ACL injury mitigation training (432), it was recommended that a mixed multicomponent programme which incorporates strength, balance, trunk control, plyometrics, and COD/agility training is performed, and these modalities would commonly feature in an athlete's holistic strength and conditioning programme. Balance training has been shown to be effective in reducing side-stepping knee joint loads (92, 93, 425), potentially attributed to favourable changes in muscle activation patterns of the hamstring, trunk, and hip musculature (92, 425). Consequently, the training intervention (Chapter 8) may have benefited from supplemental balance and dynamic trunk stability exercises and structured resistance training to help improve muscular activation and postural control. As such, future work is needed which determines the effects of a mixed multicomponent training on COD biomechanics to increase the ecological validity to the field and "real-world" environments.

Importantly, a significant strength of Chapter 8 was that changes in cutting techniques and mechanics that were associated with improved performance and reduced knee joint loads were explored, providing a novel insight into the biomechanical effects of COD technique modification. Specifically, improvements in performance were attributed to increases in velocity at key instances of the COD, increases in propulsive force production over shorter GCTs, and decreased knee flexion ROM. From an injury risk perspective, reductions in knee joint loads were primarily attributed to changes in peak and IC KAA and approach velocity. Based on the aforementioned associations, practitioners may consider implementing training strategies that enhance their athletes' speed, ability to rapidly apply force over shorter GCTs, and potentially coaching "stiffer" cutting strategies. Additionally, based on the results of Chapter 5 and 8, reducing peak and initial KAAs is a recommended deficit that practitioners should address due to its association with peak KAMs and offering no associated performance benefits. Going forwards, practitioners should pre-screen athletes prior to technique modification training to enable individual technique deficits to be targeted during the intervention, and they should consider implementing external verbal coaching cues and individual feedback when coaching COD speed training in their training programmes with multidirectional athletes.

Upon reflection, changing direction will impose multiplanar knee joint loading, and this will be inevitable when athletes become faster (the ultimate aim of strength and conditioning). Athletes will ultimately seek to adopt cutting strategies that enhance performance, even at the expense of greater knee joint loading, and it is not feasible to instruct athletes to perform CODs slower at reduced approach velocities (148). As such, to help better prepare athletes to tolerate and support the knee joint loads during COD, especially as they get faster, in addition to COD technique modification, it is recommended that practitioners develop their athletes' physical capacity through a mixed multicomponent model (i.e., neuromuscular control, co-contraction, and rapid force production) (39, 138, 272, 301, 328, 341, 410, 432, 544, 591) as highlighted in the scoping review in Chapter 2.3.5. Table 9.3 highlights physical attributes which should be targeted to assist in the physical preparation for athletes to address "high-risk" deficits associated with moment arm distance and GRF (301); thus, joint moment and ACL strain (26, 424, 512, 513), and to increase muscle activation and strength to support knee joint loads and potentially unload the ligament (138, 328, 340, 341). This should help assist and alleviate the "performance-injury conflict" during cutting.

Table. 9.1. Physical preparation recommendations

Physical qualities to develop	Rationale
Dynamic trunk control (138, 141, 292)	Because of its effect on frontal plane moment arm distance and GRF vector orientation, thus, knee joints loads during COD (141, 245). Additionally, deficits in trunk control are associated with ACL injury (635, 636).
Pelvic control (292, 534)	Because pelvic obliquity can directly influence lateral trunk flexion angle (534).
External hip rotator (gluteal) strength and activation (287, 292)	To reduce KAA thus, frontal plane moment arm and KAM (242, 359), oppose knee valgus moments and rotator moments (341), and to facilitate effective propulsion (340).
Hamstring strength and activation (31, 507, 525, 637)	To reduce ACL loading (591), enable greater knee flexion (138), and reduce anterior tibial shear and impact GRF (328, 341), while assisting in braking and propulsion (37, 340) (particularly medial hamstring strength and activation).
Quadricep strength and co-activation (37, 138, 328, 340)	Facilitate effective braking (eccentric) and propulsion (concentric) (37, 138, 328, 340), and assist knee joint stability.
Soleus strength and activation (2, 340, 341, 373, 454, 522)	To oppose anterior tibial translation and facilitate braking and propulsion (2, 340, 341, 373, 454, 522).

Key: COD: Change of direction; GRF: Ground reaction force; ACL: Anterior cruciate ligament; KAA: Knee abduction angle

9.2 Limitations, considerations, and recommended future directions of research

Limitations for each study have been presented in their respective chapters; thus, the main limitations and considerations of the thesis will be discussed in this section.

The primary angle investigated in the thesis for the biomechanical studies was 70-90˚ (intended 90˚), except for Chapter 8 where a 45˚ COD was also examined. As the biomechanical demands for COD are angle- and velocity-dependent (39, 108, 148, 212, 213, 227, 229, 504, 505, 516), the results of this study are applicable to these angles only. Additionally, side-step cutting actions were only examined. Thus, caution is advised extrapolating the results from this thesis to different COD angles and actions. As athletes may perform a range of different COD actions (XOCs, split-steps, pivots) in sport which are important for performance and knee and ankle injury mitigation, practitioners are interested in coaching guidelines to optimise performance and reduce injury risk during the COD actions. Thus, further insight is required into biomechanical determinants of performance and injury risk during different COD actions and angles to better understand optimal COD techniques.

The thesis only examined pre-planned COD biomechanics. Flashing lights/arrows have been used in other laboratories to examine unanticipated COD biomechanics but this does not provide a sport-specific stimulus. Although examining COD biomechanics in response to sport-specific stimuli would provide greater ecological validity, it is difficult to provide a sport-specific stimulus in a laboratory environment which is controlled and standardised across trials and participants. Furthermore, due to laboratory configuration, biomechanical studies in the present thesis only examined side-stepping cutting performance from the athletes' right limb, which in most cases was their preferred kicking and push-off limb. Athletes can display performance deficits and subtle differences in COD biomechanics between limbs, as highlighted in a recently published systematic review (Appendix 8.1) (145). Thus, it was unfortunate that the present thesis did not examine the biomechanical determinants of performance and injury risk for the left limb and monitor changes in cutting biomechanics for the left limb during the training intervention (Chapter 8). Feedback from athletes in the IG following the training intervention (Chapter 8) included subjectively feeling faster with the left-limb push-off. As such, future research should investigate COD for both dominant and non-dominant push-off limbs to better understand why between-limb differences during COD occur.

Although a standardised surface and footwear was used for all biomechanical studies in the thesis, this surface does not reflect the grass and artificial field-turfs that the athletes perform their CODs on. Different surfaces have different frictional properties which can influence COD biomechanics (151, 152, 326, 448) and completion time (448), and it was not feasible to perform the CODs with sport-specific footwear (i.e., studded football/rugby boots), where differences in shoe-surface interface can also alter COD biomechanics (195, 326, 380, 545) and completion times (619). Future research should compare COD biomechanics between surfaces and footwear and determine if athletes consistently display abnormal/safe and "high-risk/low-risk" mechanics between surfaces and determine if the technical and mechanical determinants are also consistent between surfaces. It must be emphasised, however, that the lack of applicability to sporting situations is one of the significant limitations of laboratory and 3D motion analysis testing methodologies; however, one of the aims of the thesis was to overcome the restricted application of cutting screening by validating a screening tool that practitioners could easily implement in the field. In Chapter 6, a qualitative screening tool has been validated against 3D motion analysis which offers a field-based method to potentially identify athletes who display "high-risk" cutting mechanics.

It is worth noting that although the results from the thesis are insightful, the findings are representative of the specific laboratory and subsequent data collection and analysis procedures. Furthermore, primarily male soccer, rugby, and cricket athletes from the United Kingdom were investigated, who were injury free and performing a pre-planned, prescriptive task. Thus, caution is advised regarding the extrapolation and generalisation of these results to different athletic populations, sexes, injured populations, and different COD tasks. Further research is needed to better understand the "performance-injury conflict" and effect of training interventions on cutting biomechanics in different athletic populations and sexes. Additionally, knee joint moments were assessed as surrogates of ACL load and potential injury risk (26, 424, 512, 513). Although numerous studies have used this approach, ACL load was not directly assessed. Musculoskeletal modelling has been recently used to estimate ACL loading during cutting (524, 591) and thus, further research is needed examining the biomechanical determinants of cutting ACL loads via musculoskeletal modelling.

In the thesis, discrete point analysis was carried out, resulting in only maximum, minimum, IC, and ROM values examined. This approach, however, can lead to regional focus bias, whereby a large amount of data are discarded from the entire waveform (385, 439-441, 490); thus, valuable information across the whole curve is unexamined because only a single data point is examined (64, 466). Additionally, discrete point analysis does not take into account the position of the key measures (i.e., differences in timing), for example, trial peaks may occur at different timings along the waveform and thus, the temporal organisation of the pattern is lost (484, 485, 538). Consequently, future research should conduct full temporal analysis of the full waveform, using statistical approaches such as SPM (191, 488, 497, 574) or temporal phase analysis (103, 363, 364). This approach could be applied, for example, when aiming to compare COD cutting biomechanics between faster and slower, "higher- and lower-risk" athletes, between-session reliability, or examining longitudinal changes in response to a training intervention. Conducting such analysis could provide further insight not only into the magnitude of differences, but importantly where these differences occur across the whole time-series. Additionally future research could consider investigating coordination (joint-joint coordination; angle-angle plots; position-velocity plots), such as the interaction between hip and knee for more insight into performance and injury risk determinants (225, 383, 597).

While positive findings regarding the validation of the CMAS against 3D motion analysis were found, a replication study performed in a different laboratory may be worthwhile to confirm the validity of the tool, and further work is required to establish the agreements and inter-rater reliability between different applied practitioners, such as sports rehabilitators, physiotherapists, and sports coaches. Additionally, future work should consider implementing the CMAS in the field by monitoring changes in CMASs over the course of the season and in response to training interventions as shown in Chapter 7, while the between-session reliability of CMAS items should also be established as previously done with TJAs (196, 476) to determine if deficits are consistently displayed between-sessions by athletes. It is worth noting that the qualitative screening is subjective and the CMAS requires ~3 minutes to screen a trial, while training and expertise is also required to identify the specific deficits (164). Consequently, future research and technology is required that can provide real-time valid and objective measures of cutting kinematics in the field.

A significant strength of the thesis was that two training intervention studies were conducted providing a novel insight into changes in performance and mechanics associated with greater knee joint loads. Short approach distances were only investigated in the training intervention (5-m); thus, the tasks are only reflective of low-entry velocity side-step cutting ability. Because both interventions focused on deceleration and braking strategies, it would have been interesting to monitor changes in deceleration performance (203, 206, 221), and sharper COD performance (i.e., 180° turns). Additionally, cutting and deceleration actions expose athletes to high eccentric loads and muscle activation of the lower-limb musculature (37, 212, 222, 340, 341, 405); therefore, it would have been interesting to monitor changes in muscle activation following the training intervention. Speculatively there may have been increased muscle activation of the hamstrings, gluteal muscles, and soleus, which may have the potential to help unload the knee ligaments (126, 328, 341). Future research should monitor muscle activation changes in addition to the biomechanical effects of COD technique modification. Additionally, Lockie et al. (332) has demonstrated positive adaptations in concentric and eccentric knee flexor and extensor strength following decelerations in combinations with COD speed training. Unfortunately, the present thesis did not monitor changes in muscle strength and is thus a recommended future direction of research.

It is worth highlighting that the changes in cutting performance and CMASs were examined within 1-week post-intervention for Chapter 7, while changes in cutting biomechanics for the IG in Chapter 8 were monitored within 2-weeks post-intervention, and only a 6-week intervention was examined. While the interventions were undoubtedly successful for enhancing performance, they were also effective in reducing CMASs and knee joint loads in "higher-risk" athletes (Chapter 7 and 8). Unfortunately, however, it is unknown whether the technique can be maintained for extensive periods and it is unclear what happens to cutting biomechanics when you discontinue this method of training. Thus, further investigation into short- and long-term retention of modified cutting technique is required to improve our understanding of training prescription and dosages. Additionally, the training interventions focused on the biomechanical effects of technique modification on surrogates of non-contact ACL injury; however, it is unknown how the technical modifications affected groin and ankle and potential injury risk. Consequently, future work is necessary that considers the biomechanical effects of cutting technique modifications on groin and ankle injury risk (surrogates of injury risk). Finally, it was evident that COD technique modification enhanced athletes' physical capacity to cut faster. Although outside the scope of the present thesis, it is unknown how this method of training impacts on-field sporting performance (i.e., tackle break success, duel success, number of evasions etc.) and unanticipated cutting mechanics, and is thus a recommended avenue for further research.

9.3 Conclusion

Overall, athletes ultimately pursue improvements in cutting performance, while minimising injury risk. Additionally, practitioners require a tool in order to screen and evaluate cutting mechanics that can be easily implemented in the field, while also understanding the cutting technical and mechanical models to coach faster and safer cutting. Finally, practitioners are interested in training interventions that improve cutting performance while mitigating ACL injury risk.

The results from the thesis bridge the gap between the laboratory and field by better understanding the biomechanical determinants of performance and injury risk during side-step cutting, validating a qualitative screening tool, and conducting two training interventions in an applied field- and laboratory setting. Critically, the results from Chapter 5 add to the body of literature regarding the biomechanical determinants of performance and injury risk, resulting in updated deterministic models and technical models for faster and safer cutting. The CMAS was validated against 3D motion analysis and offers practitioners a field-based method to identify athletes who display "high-risk" mechanics and multiplanar knee joint loads (Chapter 6). The CMAS criteria can also be used as a technical model for coaching safer cutting mechanics, where cutting technical models and guidelines are sparse, though practitioners should be conscious of the "performance-injury conflict" when modifying specific "high-risk" deficits. Finally, COD technique modification training was able to improve cutting performance and movement quality in male youth soccer players (Chapter 7), while improving cutting performance in multidirectional athletes and reducing knee joint loads in athletes considered "higher-risk" (Chapter 8). Because of the enhanced cutting performance as a result of training intervention, athletes and coaches are likely to "buy-in" and adhere to this form of training.

Ultimately, the results of the thesis highlight that an individualised and multicomponent model is needed to improve COD performance and reduce injury risk. Going forward, practitioners should initially pre-screen athletes and can consider using the CMAS to identify specific deficits and target "high-risk" athletes. The information obtained during CMAS testing can be used to help inform COD technique modification training which is an effective method for improving performance and reducing knee joint loads in "higher-risk" athletes. However, COD technique modification is just one training strategy which should be embedded into a mixed multicomponent model. Supplemental conditioning (i.e., dynamic trunk stability, balance, plyometric, strength training) is also necessary to improve an athlete's physical capacity to help alleviate the "performance-injury conflict" (Table 9.3). Overall, the deterministic and technical models, CMAS screening tool, and field-based training interventions established in the thesis are not only applicable for strength and conditioning practitioners working with healthy, injury-free athletes, but have important implications from a rehabilitative context that can be used to assist in the rehabilitation, movement profiling, and return to play of injured athletes.

9.4 Recommendations

Practical applications

- CMAS is a valid and reliable screening tool for evaluating side-step cutting movement quality which offers practitioners a cost-effective and easily applicable field-based screening tool to identify athletes who generate high peak KAMs and display aberrant mechanics.
- Techniques and mechanics associated with faster cutting performance are in direct conflict with safer cutting mechanics (i.e., reduced knee joint loading) and support the concept of a "performance-injury conflict".
- COD technique modification training improves cutting performance and reduces multiplanar knee joint loads in "higher-risk" multidirectional athletes.
- Practitioners should initially pre-screen athletes using the CMAS prior to technique modification training to enable individual technique deficits to be targeted during the intervention. Supplemental conditioning using a mixed multicomponent programme is also required to help improve an athlete's physical capacity in order to tolerate and support knee joint loading, thus ligament unloading.
- Coaching PFC dominant braking strategies and minimising knee valgus and lateral trunk flexion should facilitate effective performance and reduce knee joint loading, and are recommended deficits to target within strength and conditioning programmes.

Future directions of research

- Investigation of the biomechanical determinants of performance of injury risk during CODs of different angles, actions, athletic populations, on sport-specific surfaces (and footwear), while investigating both limbs.
- Validation of qualitative screening tools for different COD actions.
- Inspection of cutting coordination and more in-depth analysis of the full waveform for biomechanical variables (i.e., SPM).
- Training interventions that initially pre-screen individuals prior to the intervention to inform a mixed multicomponent programme which is more ecologically valid to applied settings.
- Monitoring of changes in muscle activation and strength in response to COD technique modification training.

APPENDIX

Appendix 1.1 Ethical approval

University of Salford
MANCHESTER

Research, Innovation and Academic Engagement Ethical Approval Panel

Research Centres Support Team
G0.3 Joule House
University of Salford
M5 4WT

T +44(0)161 295 2280

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1 November 2016

Dear Thomas Dos'Santos,

RE: ETHICS APPLICATION HSR1617-02 – Biomechanical determinants of injury risk and performance during change of directions: implications for screening and intervention

Based on the information you provided, I am pleased to inform you that application HSR1617-02 has been approved.

If there are any changes to the project and/ or its methodology, please inform the Panel as soon as possible by contacting Health-ResearchEthics@salford.ac.uk

Yours sincerely,

Sue McAndrew
Chair of the Research Ethics Panel

Research, Innovation and Academic Engagement Ethical Approval Panel

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M5 4WT

T +44(0)161 295 2280

www.salford.ac.uk/

22 June 2017

Dear Thomas,

RE: ETHICS APPLICATION–HSR1617-131–'Biomechanical effects of change of direction technique modification.'

Based on the information you provided I am pleased to inform you that application HSR1617-131 has been approved.

If there are any changes to the project and/or its methodology, then please inform the Panel as soon as possible by contacting Health-ResearchEthics@salford.ac.uk

Yours sincerely,

Sue McAndrew
Chair of the Research Ethics Panel

Amendment Notification Form

Please complete this form and submit it to the Health Research Ethics Panel that reviewed the original proposal: Health-ResearchEthics@Salford.ac.uk

Title of Project:
Biomechanical effects of change of direction technique modification

Name of Lead Applicant: Thomas Dos'Santos

School: School of health sciences

Are you the original Principal Investigator (PI) for this study? ***YES*** *(delete as appropriate)*

If you have selected 'NO', please explain why you are applying for the amendment:

Date when original approval was obtained: 22/6/17

Reference No: HSR1617-131

Please outline the proposed changes to the project. ***NB.*** *If the changes require any amendments to the PIS, Consent Form(s) or recruitment material, then please submit these with this form highlighting where the changes have been made:*

In light of receiving a doctoral research grant from the National Strength and Conditioning Association for $4108.00, please find attached a revised ethics application (changes highlighted in yellow), whereby participants will now receive payment for testing and adhering to a training intervention. This should help facilitate subject recruitment and minimise drop out rates, thus increasing statistical power.

Please say whether the proposed changes present any new ethical issues or changes to ethical issues that were identified in the original ethics review, and provide details of how these will be addressed:

All participants will be provided a fair opportunity to take part in the training intervention; however, participants will be chosen randomly for the control and intervention groups. Participants financial status regarding grant, bursaries, visa status to ensure payment is not affected.

Amendment Approved: ✓

Version 2.0 – 27 June 2018

Date of Approval: ***6th July 2018***

Chair's Signature:

Appendix 2.1 Effect of Low-Pass Filtering Cut-Off Frequencies on Joint Moments During Cutting

Objective: A contentious issue in the computation of joint moments, during high impact movements, is the selection of appropriate filtering COFs for marker and force data, and whether the COF should be matched or mismatched. The aim of this study was to determine the effect of different low-pass filtering combinations of matched and mismatched COFs on HFMs, KFMs, and ADFMs during cutting over the PFC and FFC, and to determine the influence of COF combination on the ranking and classification of athletes based on their KAM, as this factor is used for injury prediction.

Study design: Repeated measures, within-subject.

Setting: Laboratory (3D motion and GRF analysis).

Participants: Twenty-two athletes from multiple sports (soccer, rugby, netball, and cricket).

Methods: Joint moments were calculated through inverse dynamics during a 90° cut, whereby marker and force data were low-pass filtered using three matched COF combinations (12-12, 15-15, and 18-18 Hz) and three mismatched COF combinations (12-25, 12-50, and 15-25 Hz).

Results: Significantly greater KAM ($p \leq 0.03$, $g = 0.35$-1.49, 17.39-123.12%) (Figure 1), PFC HFMs ($p \leq 0.025$, $g = 0.68$-2.35, 25.66-143.24%), and FFC HFMs ($p \leq 0.015$, $g = 0.34$-1.08, 6.21-52.32%) were produced with mismatched COFs compared to matched COFs (Figures 1 & 2, Table 1). However, the increase in magnitude for the aforementioned variables was not always consistent across participants with mismatched COFs (Figure 2). The COF combination had minimal and no meaningful effect ($g \leq 0.23$, $\leq$ 4.14%) on PFC and FFC KFMs and ADFMs (Table 1). Participant KAM rankings were also affected by COF combination ($\rho = 0.287$-0.980) (Table 2).

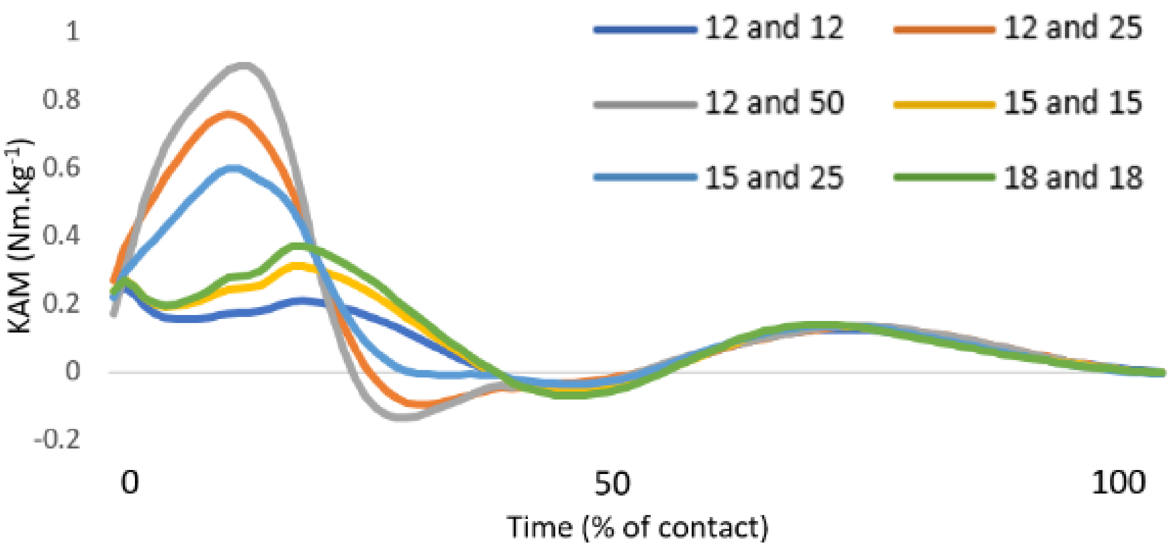


Figure 1. Comparison of KAM waveforms between matched and mismatched COFs

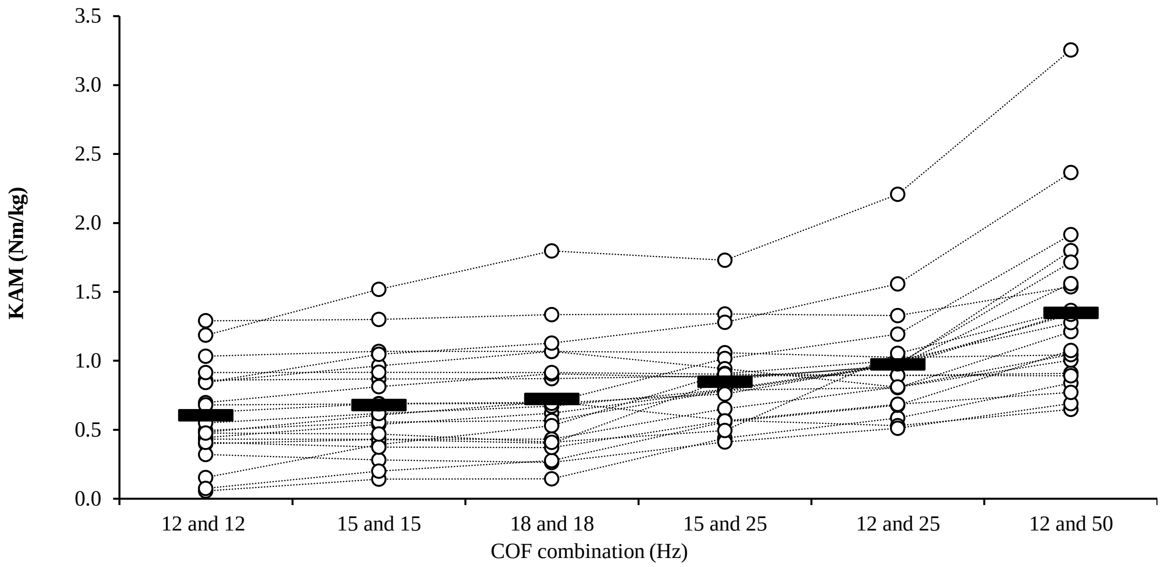


Figure 2. Individual FFC KAMs

Table 1. Repeated measures ANOVA for external joint moments between different combinations of cut-off frequencies for filtering of marker and force data

	Matched COF						Mismatched COF						
(Nm/kg)	12-12 Hz		15-15 Hz		18-18 Hz		15-25 Hz		12-25 Hz		12-50 Hz		RMANOVA
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	*p* value
KAM	0.60	0.33	0.68	0.35	0.72	0.39	0.85	0.31	0.97	0.37	1.35	0.61	<0.001
FFC HFM	-1.92	0.62	-2.04	0.68	-2.17	0.78	-2.31	0.86	-2.45	0.99	-2.93	1.14	<0.001
PFC HFM	-1.31	0.53	-1.37	0.43	-1.59	0.50	-2.00	0.66	-2.24	0.71	-3.18	0.97	<0.001
FFC KFM	3.20	0.68	3.20	0.68	3.21	0.67	3.21	0.67	3.22	0.66	3.25	0.65	0.101
PFC KFM	3.41	0.57	3.45	0.59	3.48	0.61	3.46	0.56	3.44	0.54	3.55	0.61	0.026
FFC ADFM	1.75	0.63	1.76	0.62	1.76	0.62	1.76	0.63	1.77	0.62	1.78	0.62	0.123
PFC ADFM	0.65	0.24	0.67	0.24	0.67	0.25	0.68	0.24	0.67	0.24	0.68	0.24	0.015

Key: KAM = Knee abduction moment; HFM = Hip flexor moment; KFM = Knee flexor moment; ADFM = Ankle dorsi-flexor moment; COF = Cut-off frequencies; PFC = Penultimate foot contact; FFC = Final foot contact

Table 2. Spearman's correlations for knee abduction moment rankings between cut-off frequencies

		12 – 12 Hz	12 – 25 Hz	12 – 50 Hz	15 – 15 Hz	15 – 25 Hz
12 – 25 Hz	ρ	0.571				
	p value	0.006				
12 – 50 Hz	ρ	0.287	.822			
	p value	0.195	<0.0001			
15 – 15 Hz	ρ	.980	.627	0.373		
	p value	<0.0001	0.002	0.087		
15 – 25 Hz	ρ	.768	.724	.527	.844	
	p value	<0.0001	<0.0001	0.012	<0.0001	
18 – 18 Hz	ρ	.937	.571	0.329	.979	.859
	p value	<0.0001	0.006	0.135	<0.0001	<0.0001

Conclusion: Low-pass filtering COF combination significantly affected the magnitude of KAM and PFC and FFC HFMs, while participant KAM rankings were also affected by COF combination. The differences in magnitudes and ranking may influence the interpretation of an athlete's biomechanical injury risk profile. COF combination had a minimal effect on knee and ankle dorsi-flexion moments; thus, less caution is warranted computing these moments during cutting.

Appendix 2.2 Average of Trial Peaks versus Peak of Average Profile: Impact on Change of Direction Biomechanics

(Published in Sports Biomechanics (142))

Objective: To compare lower-limb kinematic and kinetic variables during a 90° cutting task between two averaging methods of obtaining discrete data: peak of average profile vs average of individual trial peaks; and to determine the effect of averaging methods on participant ranking of each variable within a group.

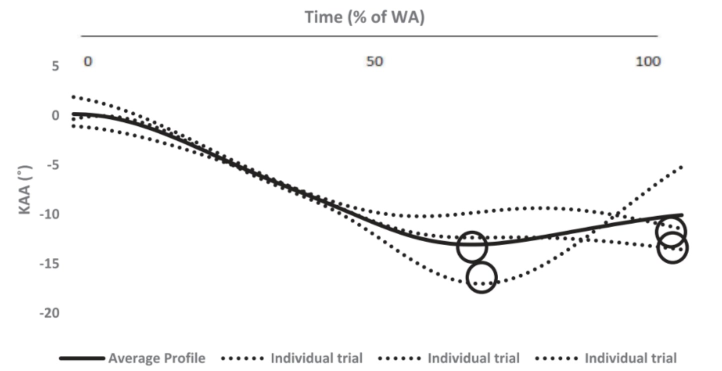


Figure 1. Example knee abduction angle (KAA) time-series data over weight acceptance (WA) illustrating variation and misalignment in peak KAA location between individual trials and average profile. (Circle denotes trial peak location.)

Design: Repeated measures, within-subject design.

Setting: Laboratory (3D motion and GRF analysis).

Participants: Twenty-two participants, from multiple sports, performed a 90˚ cut, whereby lower-limb kinematics and kinetics were assessed.

Results: Six of the eight dependent variables (VGRF and HGRF; peak HFM, KFM, and KAM, and KAA) were significantly greater ($p \leq 0.001$, g = 0.10-0.37, 2.74-10.40%) when expressed as an average of trial peaks compared to peak of average profiles (Table 1). Trivial ($g \leq 0.04$) and minimal differences (≤ 0.94%) were observed in peak hip and knee flexion angle between averaging methods (Table 1). Very strong correlations ($\rho \geq 0.901$, $p < 0.001$) were observed for rankings of participants between averaging methods for all variables.

Table 1. Comparisons in dependent variables between averaging methods.

	Average of trial peaks		Peak of average profile				Mean difference (Bias)		95% LOA		% difference			
Variable	Mean	SD	Mean	SD	p	g	Mean	SD	LB	UB	Mean	SD	ICC	SEM
pk VGRF (N/BW)	2.55	0.44	2.39	0.40	<0.001	0.37	0.15	0.13	−0.11	0.42	5.93	5.12	0957	0.10
pk HGRF (N/BW)	−1.41	0.27	−1.33	0.25	<0.001	−0.30	−0.11	0.02	−0.21	0.05	5.58	4.08	0.933	0.08
pk HFA (°)	47.52	10.0	47.07	9.95	0.001	0.04	0.45	0.54	−0.6	1.51	0.94	1.10	0.962	2.02
pk HFM (Nm/kg)	−2.65	0.91	−2.46	0.90	0.001	0.21	0.19	0.23	−0.63	0.25	7.31	8.37	0.941	0.23
pk KFA (°)	59.77	6.44	59.73	6.50	0.279	0.01	0.04	0.19	−0.33	0.42	0.09	0.38	0.863	2.64
pk KFM (Nm/kg)	3.46	0.63	3.37	0.63	<0.001	0.15	0.09	0.09	−0.09	0.28	2.74	2.69	0.936	0.17
pk KAA (°)	−8.16	8.09	−7.31	8.09	<0.001	−0.10	−0.86	0.92	−2.66	0.95	10.40	9.41	0.982	1.10
pk KAM (Nm/kg)	0.99	0.36	0.92	0.34	<0.001	0.20	0.07	0.07	−0.07	0.21	7.03	6.74	0.909	0.12

pk, peak; VGRF, vertical ground reaction force; HGRF, horizontal ground reaction force; HFA, hip flexion angle; HFM, hip flexor moment; KFA, knee flexion angle; KFM, knee flexor moment; KAA, knee abduction angle; KAM, knee abduction moment; LOA, limits of agreement; LB, lower bound; UB, upper bound; BW, body weight; ICC, intraclass correlation coefficient; SEM, standard error of measurement.

Conclusion: Practitioners and researchers should obtain discrete data based on the average of trial peaks because it is not influenced by misalignments and variations in trial peak locations, in contrast to the peak from average profile (Figure 1).

Appendix 3.1 Within-session reliability measures (Chapter 4)

90˚ cut - session 1 within-session reliability measures

Foot contact	Variable	Trial size 3 (1-3)					Trial size 5 (1-5)					Trial size 8 (1-8)					Trial size 10 (1-10)				
		Mean	SD	ICC	CV%	SEM	Mean	SD	ICC	CV%	SEM	Mean	SD	ICC	CV%	SEM	Mean	SD	ICC	CV%	SEM
FFC	PK ADFA (˚)	74.8	8.4	0.950	4.3	2.0	74.9	7.8	0.951	4.6	1.9	75.8	7.7	0.961	5.1	1.7	76.1	8.1	0.968	5.7	1.7
	PK KFA (˚)	58.6	5.4	0.783	6.5	3.1	58.3	5.5	0.870	6.8	2.5	57.8	5.3	0.911	7.6	2.0	57.8	5.3	0.930	7.6	1.8
	PK HFA (˚)	46.6	8.6	0.900	9.8	3.0	45.7	9.2	0.954	9.7	2.1	44.5	9.4	0.972	10.1	1.7	44.2	9.2	0.978	9.9	1.5
	PK ADFM (Nm/kg)	1.67	0.37	0.811	14.7	0.19	1.68	0.35	0.814	**18.4**	0.20	1.72	0.37	0.894	**18.6**	0.16	1.76	0.37	0.908	**19.1**	0.15
	PK KFM (Nm/kg)	3.45	0.59	0.898	8.3	0.21	3.42	0.57	0.932	9.4	0.17	3.42	0.50	0.951	9.0	0.13	3.42	0.49	0.961	9.0	0.11
	PK HFM (Nm/kg)	2.51	0.76	0.923	**15.8**	0.22	2.45	0.78	0.945	**19.3**	0.20	2.44	0.80	0.971	**18.6**	0.15	2.47	0.83	0.979	**17.9**	0.13
	PK KAA (˚)	7.3	6.3	0.970	**49.5**	1.1	7.9	6.6	0.980	**41.8**	1.0	8.2	6.8	0.989	**38.5**	0.7	8.1	6.8	0.991	**39.1**	0.7
	PK KAM (Nm/kg)	1.04	0.31	0.758	**25.0**	0.19	1.07	0.36	0.902	**23.7**	0.13	1.09	0.38	0.948	**23.6**	0.10	1.08	0.41	0.965	**24.3**	0.09
	PK KIRM (Nm/kg)	-0.81	0.28	0.786	**22.7**	0.15	-0.80	0.26	0.777	**29.7**	0.17	-0.83	0.31	0.890	**29.4**	0.14	-0.83	0.32	0.912	**30.2**	0.13
	Mean HBF (BW)	0.83	0.13	0.903	8.1	0.04	0.84	0.14	0.931	9.5	0.04	0.85	0.12	0.950	8.8	0.03	0.86	0.12	0.953	9.1	0.03
	Mean VBF (BW)	1.61	0.24	0.928	6.2	0.07	1.61	0.24	0.950	6.8	0.06	1.62	0.23	0.972	6.3	0.04	1.63	0.23	0.977	6.6	0.04
	PK HBF (BW)	1.42	0.42	0.902	13.4	0.14	1.43	0.42	0.957	12.5	0.09	1.42	0.39	0.964	13.1	0.08	1.43	0.39	0.972	13.0	0.07
	PK VBF (BW)	2.65	0.64	0.869	12.0	0.26	2.62	0.62	0.929	11.6	0.19	2.60	0.59	0.954	12.2	0.15	2.63	0.61	0.964	12.7	0.13
PFC	PK ADFA (˚)	78.9	8.0	0.875	5.0	3.2	78.8	7.8	0.902	6.3	2.9	79.4	7.4	0.931	6.6	2.4	79.1	7.9	0.945	7.0	2.3
	PK KFA (˚)	104.8	9.3	0.896	4.4	3.3	105.0	9.1	0.939	4.4	2.5	105.2	9.2	0.964	4.5	2.0	104.7	9.2	0.971	4.6	1.8
	PK HFA (˚)	55.4	6.7	0.876	5.5	2.7	55.8	6.5	0.930	6.2	1.9	56.0	6.2	0.953	6.6	1.6	55.9	6.3	0.959	7.1	1.5
	PK ADFM (Nm/kg)	0.69	0.23	0.732	**26.2**	0.15	0.69	0.24	0.874	**26.2**	0.11	0.70	0.24	0.936	**24.4**	0.07	0.70	0.25	0.946	**25.9**	0.07
	PK KFM (Nm/kg)	3.66	0.68	0.906	8.4	0.23	3.60	0.74	0.944	10.6	0.19	3.58	0.73	0.962	11.2	0.16	3.58	0.72	0.966	12.0	0.15
	PK HFM (Nm/kg)	1.59	0.71	0.912	**26.0**	0.23	1.60	0.69	0.958	**21.5**	0.15	1.62	0.67	0.967	**22.0**	0.14	1.58	0.63	0.964	**25.3**	0.14
	Mean HBF (BW)	0.56	0.15	0.943	10.4	0.04	0.56	0.15	0.971	9.9	0.03	0.56	0.15	0.982	10.4	0.02	0.56	0.15	0.985	11.0	0.02
	Mean VBF (BW)	0.98	0.22	0.976	6.0	0.04	0.97	0.22	0.978	7.5	0.03	0.97	0.23	0.985	8.2	0.03	0.97	0.23	0.988	8.4	0.03
	PK HBF (BW)	1.41	0.50	0.941	**15.2**	0.13	1.43	0.50	0.955	**17.5**	0.11	1.44	0.52	0.977	**17.0**	0.09	1.46	0.52	0.979	**17.7**	0.08
	PK VBF (BW)	2.35	0.59	0.941	9.9	0.15	2.37	0.58	0.952	11.8	0.14	2.37	0.58	0.973	11.8	0.10	2.41	0.57	0.978	11.4	0.09
GCT	FFC GCT (s)	0.292	0.033	0.793	7.6	0.018	0.291	0.033	0.881	7.4	0.014	0.286	0.028	0.885	8.3	0.012	0.284	0.025	0.891	8.4	0.012
	PFC GCT (s)	0.213	0.030	0.824	8.9	0.014	0.212	0.028	0.813	10.7	0.016	0.211	0.029	0.902	10.7	0.012	0.208	0.028	0.922	10.8	0.010
		Average		0.884	12.8				0.926	13.1				0.952	13.0				0.960	13.5	

Key: ADFM: Ankle dorsi-flexion moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFA: Ankle dorsi-flexion angle; KFA: Knee flexion angle; HFA: Hip flexion angle; FFC: Final foot contact; PFC: Penultimate foot contact; PK: Peak; GRF: Ground reaction force; KAA: Knee abduction angle; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; HBF: Horizontal braking force; VBF: Vertical braking force; GCT: Ground contact time; WA: Weight acceptance; ICC: Intraclass correlation coefficient; CV%: Coefficient of variation; SD: Standard deviation; SEM: Standard error of mean. Bold denotes exceeds acceptable reliability criteria

90˚ cut - session 2 within-session reliability measures

Foot contact	Variable	Trial size 3 (1-3)					Trial size 5 (1-5)					Trial size 8 (1-8)					Trial size 10 (1-10)				
		Mean	**SD**	**ICC**	**CV%**	**SEM**	**Mean**	**SD**	**ICC**	**CV%**	**SEM**	**Mean**	**SD**	**ICC**	**CV%**	**SEM**	**Mean**	**SD**	**ICC**	**CV%**	**SEM**
FFC	PK ADFA (˚)	76.8	8.3	0.938	4.1	2.2	76.2	8.7	0.960	4.8	1.9	76.1	9.0	0.979	4.8	1.4	76.1	8.8	0.982	4.8	1.3
	PK KFA (˚)	60.3	8.0	0.974	3.5	1.3	59.9	8.0	0.962	5.5	1.7	59.8	7.9	0.969	6.5	1.5	59.4	7.7	0.970	7.1	1.5
	PK HFA (˚)	43.5	12.1	0.987	5.6	1.4	43.7	11.9	0.988	7.1	1.3	43.2	11.8	0.991	7.7	1.2	42.9	11.7	0.993	7.9	1.0
	PK ADFM (Nm/kg)	1.76	0.33	0.867	12.0	0.14	1.71	0.36	0.799	**17.6**	0.22	1.75	0.33	0.843	**19.7**	0.19	1.76	0.32	0.868	**19.9**	0.17
	PK KFM (Nm/kg)	3.66	0.52	0.924	5.6	0.15	3.62	0.52	0.937	6.9	0.15	3.63	0.49	0.954	7.6	0.12	3.60	0.49	0.963	7.8	0.11
	PK HFM (Nm/kg)	2.19	0.65	0.917	14.5	0.20	2.24	0.64	0.946	**15.2**	0.17	2.25	0.63	0.959	**16.8**	0.15	2.28	0.61	0.961	**17.3**	0.14
	PK KAA (˚)	5.7	5.5	0.956	**46.4**	1.2	6.2	5.8	0.973	**48.6**	1.0	6.3	5.9	0.986	**46.7**	0.7	6.5	5.9	0.989	**46.6**	0.7
	PK KAM (Nm/kg)	0.88	0.40	0.927	**22.7**	0.12	0.91	0.40	0.938	**24.0**	0.11	0.89	0.39	0.962	**23.3**	0.09	0.91	0.38	0.972	**22.4**	0.07
	PK KIRM (Nm/kg)	-0.67	0.22	0.731	**22.5**	0.15	-0.73	0.28	0.810	**28.2**	0.16	-0.76	0.30	0.914	**26.2**	0.11	-0.76	0.28	0.917	**28.0**	0.11
	Mean HBF (BW)	0.90	0.22	0.979	5.9	0.03	0.90	0.19	0.962	8.5	0.04	0.90	0.18	0.968	9.0	0.04	0.91	0.17	0.974	9.0	0.03
	Mean VBF (BW)	1.67	0.36	0.992	3.0	0.03	1.67	0.34	0.990	4.5	0.04	1.66	0.32	0.986	6.0	0.04	1.67	0.32	0.990	5.7	0.03
	PK HBF (BW)	1.42	0.51	0.956	9.7	0.11	1.44	0.48	0.978	10.1	0.08	1.44	0.46	0.981	11.5	0.07	1.45	0.45	0.984	11.9	0.06
	PK VBF (BW)	2.54	0.76	0.977	6.2	0.12	2.53	0.72	0.986	7.4	0.09	2.52	0.70	0.981	9.7	0.10	2.53	0.68	0.986	9.6	0.09
PFC	PK ADFA (˚)	80.8	10.9	0.875	4.9	4.1	80.2	9.7	0.902	5.4	3.3	79.6	9.2	0.931	5.8	2.7	79.1	8.9	0.945	6.4	2.4
	PK KFA (˚)	103.3	11.8	0.936	4.2	3.2	103.7	12.5	0.975	4.0	2.1	103.5	12.1	0.985	3.8	1.6	103.5	11.9	0.987	3.9	1.4
	PK HFA (˚)	55.4	6.5	0.902	5.5	2.2	55.6	6.9	0.961	5.3	1.5	55.8	6.9	0.971	6.0	1.3	55.8	6.7	0.974	6.2	1.2
	PK ADFM (Nm/kg)	0.73	0.28	0.902	**16.4**	0.10	0.74	0.27	0.933	**19.5**	0.08	0.74	0.25	0.949	**22.7**	0.07	0.72	0.24	0.958	**22.8**	0.06
	PK KFM (Nm/kg)	3.41	0.64	0.978	4.8	0.10	3.44	0.64	0.969	7.2	0.12	3.43	0.68	0.979	8.2	0.11	3.45	0.67	0.982	8.2	0.10
	PK HFM (Nm/kg)	1.86	0.99	0.962	**20.2**	0.20	1.86	0.92	0.971	**21.4**	0.17	1.85	0.84	0.979	**22.1**	0.13	1.82	0.79	0.977	**23.6**	0.13
	Mean HBF (BW)	0.55	0.13	0.952	9.0	0.03	0.55	0.14	0.979	8.5	0.02	0.55	0.15	0.986	9.1	0.02	0.56	0.15	0.988	9.3	0.02
	Mean VBF (BW)	0.96	0.22	0.940	9.6	0.06	0.96	0.23	0.973	8.7	0.04	0.96	0.23	0.986	8.1	0.03	0.95	0.23	0.988	8.2	0.03
	PK HBF (BW)	1.36	0.47	0.964	10.8	0.09	1.35	0.49	0.979	12.0	0.07	1.36	0.51	0.984	13.7	0.07	1.37	0.52	0.988	13.3	0.06
	PK VBF (BW)	2.38	0.61	0.961	7.4	0.13	2.35	0.59	0.970	9.3	0.11	2.34	0.63	0.984	10.0	0.08	2.35	0.64	0.988	9.7	0.07
GCT	FFC GCT (s)	0.279	0.029	0.944	4.1	0.007	0.280	0.029	0.919	6.3	0.009	0.278	0.031	0.954	6.6	0.008	0.277	0.030	0.962	6.5	0.007
	PFC GCT (s)	0.194	0.026	0.763	9.8	0.016	0.196	0.028	0.912	8.5	0.010	0.193	0.025	0.934	9.1	0.008	0.192	0.024	0.942	9.2	0.007
		Average		0.936	10.2	0.936			0.953	11.5				0.966	12.3				0.971	12.4	

Key: ADFM: Ankle dorsi-flexion moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFA: Ankle dorsi-flexion angle; KFA: Knee flexion angle; HFA: Hip flexion angle; FFC: Final foot contact; PFC: Penultimate foot contact; PK: Peak; GRF: Ground reaction force; KAA: Knee abduction angle; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; HBF: Horizontal braking force; VBF: Vertical braking force; GCT: Ground contact time; WA: Weight acceptance; ICC: Intraclass correlation coefficient; CV%: Coefficient of variation; SD: Standard deviation; SEM: Standard error of mean. Bold denotes exceeds acceptable reliability criteria

90° cut - Session 1 within-session ICC and CV% with 95% CIs

Foot contact	Variable	Trial size 3 (1-3)						Trial size 5 (1-5)						Trial size 8 (1-8)						Trial size 10 (1-10)					
		ICC	95% CI LB	95% CI UB	CV%	95% CI LB	95% CI UB	ICC	95% CI LB	95% CI UB	CV%	95% CI LB	95% CI UB	ICC	95% CI LB	95% CI UB	CV%	95% CI LB	95% CI UB	ICC	95% CI LB	95% CI UB	CV%	95% CI LB	95% CI UB
FFC	PK ADFA	0.950	0.852	0.986	4.3	3.4	5.1	0.951	0.879	0.986	4.6	3.3	5.9	0.961	0.911	0.989	5.1	3.5	6.8	0.968	0.927	0.991	5.7	4.4	7.0
	PK KFA	0.783	0.334	0.942	6.5	3.7	9.4	0.870	0.675	0.963	6.8	3.9	9.6	0.911	0.794	0.974	7.6	5.6	9.6	0.930	0.842	0.980	7.6	5.8	9.3
	PK HFA	0.900	0.717	0.973	9.8	4.9	14.7	0.954	0.886	0.987	9.7	5.1	14.2	0.972	0.934	0.992	10.1	6.3	13.9	0.978	0.949	0.993	9.9	6.4	13.4
	PK ADFM	0.811	0.465	0.948	14.7	10.2	19.3	0.814	0.542	0.947	18.4	12.7	24.2	0.894	0.756	0.969	18.6	13.5	23.7	0.908	0.793	0.973	19.1	13.6	24.6
	PK KFM	0.898	0.704	0.972	8.3	4.6	11.9	0.932	0.831	0.981	9.4	6.6	12.2	0.951	0.887	0.986	9.0	6.8	11.1	0.961	0.911	0.989	9.0	7.2	10.8
	PK HFM	0.923	0.772	0.979	15.8	9.0	22.6	0.945	0.866	0.984	19.3	9.6	28.9	0.971	0.934	0.992	18.6	10.1	27.1	0.979	0.952	0.994	17.9	9.9	26.0
	PK KAA	0.970	0.911	0.992	49.5	22.2	76.9	0.980	0.952	0.994	41.8	23.7	59.9	0.989	0.974	0.997	38.5	24.0	53.1	0.991	0.979	0.997	39.1	25.0	53.2
	PK KAM	0.758	0.266	0.935	25.0	15.4	34.7	0.902	0.756	0.972	23.7	17.2	30.1	0.948	0.880	0.985	23.6	19.1	28.0	0.965	0.921	0.990	24.3	19.0	29.7
	PK KIRM	0.786	0.395	0.942	22.7	13.2	32.2	0.777	0.438	0.937	29.7	23.0	36.4	0.890	0.745	0.968	29.4	24.6	34.2	0.912	0.801	0.974	30.2	25.0	35.4
	Mean HBF	0.903	0.722	0.974	8.1	5.0	11.1	0.931	0.832	0.980	9.5	6.4	12.6	0.950	0.885	0.985	8.8	6.0	11.5	0.953	0.894	0.986	9.1	6.5	11.6
	Mean VBF	0.928	0.792	0.980	6.2	3.7	8.6	0.950	0.878	0.986	6.8	3.9	9.6	0.972	0.935	0.992	6.3	4.2	8.4	0.977	0.949	0.993	6.6	5.0	8.2
	PK HBF	0.902	0.724	0.973	13.4	7.0	19.9	0.957	0.894	0.988	12.5	7.9	17.1	0.964	0.917	0.990	13.1	9.1	17.1	0.972	0.936	0.992	13.0	9.9	16.2
	PK VBF	0.869	0.620	0.965	12.0	6.5	17.4	0.929	0.824	0.980	11.6	6.6	16.6	0.954	0.894	0.987	12.2	8.5	16.0	0.964	0.918	0.989	12.7	9.8	15.6
PFC	PK ADFA	0.875	0.625	0.966	5.0	3.1	6.9	0.902	0.754	0.972	6.3	4.7	7.9	0.931	0.841	0.980	6.6	5.3	7.8	0.945	0.874	0.984	7.0	5.4	8.6
	PK KFA	0.896	0.694	0.972	4.4	2.5	6.3	0.939	0.848	0.983	4.4	2.9	6.0	0.964	0.918	0.990	4.5	3.2	5.7	0.971	0.936	0.992	4.6	3.5	5.6
	PK HFA	0.876	0.637	0.966	5.5	1.8	9.1	0.930	0.827	0.980	6.2	3.9	8.5	0.953	0.892	0.986	6.6	5.2	7.9	0.959	0.907	0.988	7.1	5.9	8.3
	PK ADFM	0.732	0.221	0.927	26.2	15.8	36.7	0.874	0.685	0.964	26.2	19.7	32.7	0.936	0.852	0.982	24.4	20.1	28.7	0.946	0.878	0.984	25.9	20.7	31.0
	PK KFM	0.906	0.728	0.974	8.4	4.7	12.1	0.944	0.862	0.984	10.6	7.1	14.1	0.962	0.912	0.989	11.2	7.9	14.4	0.966	0.923	0.990	12.0	8.9	15.1
	PK HFM	0.912	0.752	0.976	26.0	17.1	34.8	0.958	0.896	0.988	21.5	16.3	26.7	0.967	0.925	0.991	22.0	16.7	27.2	0.964	0.919	0.990	25.3	19.9	30.7
	Mean HBF	0.943	0.837	0.984	10.4	5.0	15.8	0.971	0.928	0.992	9.9	6.5	13.3	0.982	0.959	0.995	10.4	7.9	13.0	0.985	0.966	0.996	11.0	8.2	13.8
	Mean VBF	0.976	0.931	0.993	6.0	3.7	8.3	0.978	0.945	0.994	7.5	6.1	8.8	0.985	0.965	0.996	8.2	6.2	10.2	0.988	0.972	0.996	8.4	6.4	10.3
	PK HBF	0.941	0.831	0.984	15.2	7.3	23.0	0.955	0.889	0.987	17.5	11.3	23.7	0.977	0.948	0.993	17.0	11.8	22.1	0.979	0.952	0.994	17.7	11.2	24.1
	PK VBF	0.941	0.831	0.984	9.9	5.0	14.7	0.952	0.882	0.986	11.8	8.0	15.5	0.973	0.938	0.992	11.8	9.2	14.5	0.978	0.950	0.994	11.4	8.3	14.5
GCT	FFC GCT	0.793	0.392	0.944	7.6	4.7	10.6	0.881	0.708	0.966	7.4	4.8	10.0	0.885	0.737	0.966	8.3	5.8	10.7	0.891	0.756	0.968	8.4	5.9	10.8
	PFC GCT	0.824	0.503	0.952	8.9	6.5	11.4	0.813	0.551	0.946	10.7	7.6	13.9	0.902	0.776	0.971	10.7	8.0	13.4	0.922	0.824	0.977	10.8	8.1	13.4

Key: ADFM: Ankle dorsi-flexion moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFA: Ankle dorsi-flexion angle; KFA: Knee flexion angle; HFA: Hip flexion angle; FFC: Final foot contact; PFC: Penultimate foot contact; PK: Peak; GRF: Ground reaction force; KAA: Knee abduction angle; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; HBF: Horizontal braking force; VBF: Vertical braking force; GCT: Ground contact time; ICC: Intraclass correlation coefficient; CV%: Coefficient of variation; CI: Confidence interval; UB: Upper bound; LB: Lower bound.

90° cut - Session 2 within-session ICC and CV% with 95% CIs

Foot contact	Variable	Trial size 3 (1-3)						Trial size 5 (1-5)						Trial size 8 (1-8)						Trial size 10 (1-10)					
		ICC	95% CI LB	95% CI UB	CV%	95% CI LB	95% CI UB	ICC	95% CI LB	95% CI UB	CV%	95% CI LB	95% CI UB	ICC	95% CI LB	95% CI UB	CV%	95% CI LB	95% CI UB	ICC	95% CI LB	95% CI UB	CV%	95% CI LB	95% CI UB
FFC	PK ADFA	0.938	0.822	0.983	4.1	2.4	5.9	0.960	0.903	0.989	4.8	3.1	6.4	0.979	0.951	0.994	4.8	3.5	6.0	0.982	0.960	0.995	4.8	3.7	5.9
	PK KFA	0.974	0.924	0.993	3.5	2.0	5.0	0.962	0.908	0.989	5.5	3.7	7.2	0.969	0.928	0.991	6.5	5.2	7.9	0.970	0.932	0.991	7.1	5.6	8.5
	PK HFA	0.987	0.963	0.997	5.6	3.2	7.9	0.988	0.969	0.997	7.1	5.0	9.1	0.991	0.980	0.997	7.7	5.5	9.8	0.993	0.983	0.998	7.9	5.7	10.0
	PK ADFM	0.867	0.620	0.964	12.0	7.7	16.2	0.799	0.501	0.943	17.6	11.0	24.3	0.843	0.636	0.954	19.7	13.9	25.6	0.868	0.699	0.961	19.9	14.6	25.1
	PK KFM	0.924	0.785	0.979	5.6	3.2	7.9	0.937	0.846	0.982	6.9	4.8	9.1	0.954	0.895	0.987	7.6	6.2	9.0	0.963	0.917	0.989	7.8	6.4	9.2
	PK HFM	0.917	0.753	0.978	14.5	8.6	20.3	0.946	0.867	0.985	15.2	11.8	18.7	0.959	0.904	0.988	16.8	11.7	21.9	0.961	0.912	0.989	17.3	12.4	22.2
	PK KAA	0.956	0.876	0.988	46.4	30.3	62.6	0.973	0.933	0.992	48.6	31.8	65.4	0.986	0.967	0.996	46.7	27.8	65.5	0.989	0.975	0.997	46.6	28.8	64.4
	PK KAM	0.927	0.784	0.980	22.7	15.8	29.6	0.938	0.847	0.982	24.0	18.3	29.7	0.962	0.913	0.989	23.3	18.2	28.3	0.972	0.936	0.992	22.4	18.1	26.6
	PK KIRM	0.731	0.470	0.900	22.5	12.6	32.4	0.810	0.54	0.95	28.2	21.6	34.8	0.914	0.802	0.975	26.2	20.8	31.6	0.917	0.81	0.98	28.0	22.9	33.0
	Mean HBF	0.979	0.941	0.994	5.9	5.0	6.7	0.962	0.906	0.989	8.5	6.2	10.8	0.968	0.926	0.991	9.0	6.8	11.2	0.974	0.942	0.992	9.0	7.2	10.8
	Mean VBF	0.992	0.978	0.998	3.0	2.0	3.9	0.990	0.974	0.997	4.5	3.6	5.5	0.986	0.968	0.996	6.0	4.6	7.3	0.990	0.976	0.997	5.7	4.5	7.0
	PK HBF	0.956	0.871	0.988	9.7	6.5	12.9	0.978	0.945	0.994	10.1	7.2	13.0	0.981	0.957	0.995	11.5	9.8	13.1	0.984	0.964	0.995	11.9	10.1	13.6
	PK VBF	0.977	0.934	0.994	6.2	3.8	8.5	0.986	0.965	0.996	7.4	5.1	9.7	0.981	0.957	0.995	9.7	7.5	11.9	0.986	0.968	0.996	9.6	8.0	11.3
PFC	PK ADFA	0.875	0.625	0.966	4.9	3.0	6.7	0.902	0.754	0.972	5.4	4.0	6.8	0.931	0.841	0.980	5.8	4.6	7.1	0.945	0.874	0.984	6.4	4.6	8.2
	PK KFA	0.936	0.819	0.983	4.2	2.3	6.1	0.975	0.939	0.993	4.0	2.8	5.1	0.985	0.966	0.996	3.8	2.9	4.7	0.987	0.971	0.996	3.9	3.1	4.8
	PK HFA	0.902	0.709	0.974	5.5	3.4	7.7	0.961	0.904	0.989	5.3	3.9	6.6	0.971	0.933	0.992	6.0	4.9	7.1	0.974	0.940	0.992	6.2	5.4	7.0
	PK ADFM	0.902	0.724	0.973	16.4	10.0	22.8	0.933	0.835	0.981	19.5	14.4	24.5	0.949	0.883	0.985	22.7	17.8	27.5	0.958	0.904	0.988	22.8	18.1	27.6
	PK KFM	0.978	0.937	0.994	4.8	3.5	6.1	0.969	0.924	0.991	7.2	6.0	8.5	0.979	0.951	0.994	8.2	6.7	9.7	0.982	0.959	0.995	8.2	6.6	9.7
	PK HFM	0.962	0.891	0.990	20.2	8.0	32.3	0.971	0.929	0.992	21.4	12.6	30.3	0.979	0.952	0.994	22.1	12.7	31.4	0.977	0.947	0.993	23.6	15.1	32.0
	Mean HBF	0.952	0.865	0.987	9.0	6.8	11.3	0.979	0.950	0.994	8.5	6.5	10.4	0.986	0.968	0.996	9.1	7.3	10.9	0.988	0.973	0.996	9.3	7.8	10.9
	Mean VBF	0.940	0.824	0.984	9.6	5.6	13.6	0.973	0.934	0.992	8.7	5.7	11.8	0.986	0.968	0.996	8.1	5.9	10.3	0.988	0.973	0.996	8.2	5.8	10.6
	PK HBF	0.964	0.896	0.990	10.8	8.2	13.4	0.979	0.947	0.994	12.0	9.0	15.0	0.984	0.963	0.995	13.7	9.8	17.5	0.988	0.974	0.997	13.3	9.9	16.7
	PK VBF	0.961	0.888	0.990	7.4	3.7	11.0	0.970	0.927	0.992	9.3	6.6	11.9	0.984	0.963	0.995	10.0	7.7	12.2	0.988	0.973	0.996	9.7	7.9	11.5
GCT	FFC GCT	0.944	0.842	0.985	4.1	3.1	5.1	0.919	0.799	0.977	6.3	4.9	7.8	0.954	0.894	0.987	6.6	5.3	7.9	0.962	0.915	0.989	6.5	5.5	7.5
	PFC GCT	0.763	0.299	0.936	9.8	6.5	13.0	0.912	0.781	0.975	8.5	6.5	10.5	0.934	0.847	0.981	9.1	7.4	10.8	0.942	0.869	0.983	9.2	7.8	10.6

Key: ADFM: Ankle dorsi-flexion moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFA: Ankle dorsi-flexion angle; KFA: Knee flexion angle; HFA: Hip flexion angle; FFC: Final foot contact; PFC: Penultimate foot contact; PK: Peak; GRF: Ground reaction force; KAA: Knee abduction angle; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; HBF: Horizontal braking force; VBF: Vertical braking force; GCT: Ground contact time; ICC: Intraclass correlation coefficient; CV%: Coefficient of variation; CI: Confidence interval; UB: Upper bound; LB: Lower bound.

Appendix 3.2 Between-session reliability measures containing confidence intervals (Chapter 4)

Between-session ICC and CV% with 95% CIs

Foot contact	Variable	Trial size 3 (1-3)						Trial size 5 (1-5)						Trial size 8 (1-8)						Trial size 10 (1-10)					
		ICC	95% CI LB	95% CI UB	CV%	95% CI LB	95% CI UB	ICC	95% CI LB	95% CI UB	CV%	95% CI LB	95% CI UB	ICC	95% CI LB	95% CI UB	CV%	95% CI LB	95% CI UB	ICC	95% CI LB	95% CI UB	CV%	95% CI LB	95% CI UB
FFC	PK ADFA	0.897	0.612	0.974	3.9	1.9	5.8	0.958	0.839	0.989	2.4	1.0	3.7	0.959	0.834	0.990	2.6	1.6	3.7	0.967	0.864	0.992	2.4	1.5	3.3
	PK KFA	0.719	-0.117	0.930	6.1	3.6	8.6	0.704	-0.186	0.927	5.3	2.2	8.4	0.750	0.063	0.937	4.8	1.9	7.8	0.787	0.181	0.947	4.0	1.1	7.0
	PK HFA	0.804	0.272	0.950	11.3	5.2	17.5	0.823	0.305	0.956	10.3	4.1	16.4	0.862	0.446	0.966	9.4	3.4	15.3	0.859	0.431	0.965	9.4	3.4	15.4
	PK ADFM	0.414	-1.558	0.857	14.2	8.7	19.7	0.707	-0.293	0.929	11.9	8.8	15.1	0.842	0.340	0.961	8.6	6.3	11.0	0.846	0.345	0.962	7.4	3.9	10.8
	PK KFM	0.911	0.500	0.979	5.7	2.6	8.8	0.900	0.535	0.976	5.5	2.6	8.4	0.918	0.301	0.983	4.5	2.1	6.9	0.932	0.452	0.986	4.0	1.7	6.3
	PK HFM	0.704	-0.032	0.924	16.5	8.0	25.0	0.727	-0.039	0.931	15.0	7.9	22.1	0.756	0.063	0.939	15.7	10.0	21.3	0.760	0.079	0.940	15.9	10.7	21.2
	PK KAA	0.915	0.665	0.979	45.0	29.2	60.8	0.929	0.697	0.983	35.6	20.4	50.8	0.927	0.674	0.982	36.0	20.7	51.3	0.938	0.729	0.985	32.1	15.5	48.6
	PK KAM	0.880	0.378	0.972	16.1	8.9	23.2	0.904	0.360	0.979	14.5	7.4	21.5	0.909	0.013	0.983	15.2	8.7	21.7	0.920	0.326	0.983	13.1	7.1	19.0
	PK KIRM	0.653	-0.154	0.909	19.9	11.0	28.8	0.778	0.156	0.944	17.2	7.6	26.8	0.899	0.620	0.974	13.9	5.6	22.2	0.892	0.593	0.973	13.7	5.2	22.2
	Mean HBF	0.872	0.428	0.969	7.2	4.1	10.3	0.903	0.518	0.977	5.9	2.9	8.9	0.892	0.545	0.973	5.4	2.4	8.4	0.905	0.560	0.977	5.3	2.9	7.7
	Mean VBF	0.927	0.724	0.981	5.6	3.2	8.0	0.947	0.796	0.987	5.2	3.5	6.8	0.952	0.817	0.988	4.6	2.9	6.3	0.963	0.858	0.991	3.9	2.4	5.5
	PK HBF	0.977	0.905	0.994	6.1	3.3	9.0	0.978	0.913	0.995	5.3	3.4	7.2	0.981	0.924	0.995	4.6	2.7	6.6	0.980	0.922	0.995	4.6	2.4	6.8
	PK VBF	0.916	0.678	0.979	8.3	3.5	13.2	0.928	0.721	0.982	7.2	2.5	11.9	0.943	0.781	0.986	6.4	2.3	10.5	0.938	0.764	0.984	7.1	3.0	11.1
PFC	PK ADFA	0.787	0.147	0.947	6.0	3.6	8.4	0.842	0.372	0.961	5.0	3.0	6.9	0.826	0.260	0.957	5.2	3.7	6.7	0.837	0.304	0.960	5.2	3.8	6.5
	PK KFA	0.926	0.718	0.982	3.4	1.8	5.0	0.955	0.829	0.989	2.6	1.3	4.0	0.961	0.849	0.990	2.6	1.5	3.6	0.960	0.847	0.990	2.6	1.7	3.5
	PK HFA	0.920	0.671	0.980	3.8	2.2	5.3	0.916	0.654	0.979	3.7	2.2	5.3	0.892	0.551	0.973	4.3	2.8	5.7	0.873	0.464	0.969	4.3	2.4	6.2
	PK ADFM	0.725	-0.153	0.932	20.5	13.1	27.8	0.872	0.511	0.968	15.0	11.5	18.4	0.905	0.637	0.976	11.2	7.0	15.3	0.925	0.696	0.981	10.0	5.9	14.2
	PK KFM	0.843	0.396	0.961	8.7	4.7	12.7	0.843	0.407	0.960	8.2	3.5	13.0	0.864	0.484	0.966	7.0	2.1	11.9	0.878	0.533	0.969	6.5	2.1	11.0
	PK HFM	0.702	-0.118	0.925	24.9	12.9	37.0	0.711	-0.078	0.927	23.1	10.9	35.3	0.780	0.180	0.944	20.9	8.5	33.4	0.792	0.238	0.947	19.9	8.1	31.8
	Mean HBF	0.929	0.710	0.982	7.0	2.8	11.2	0.929	0.714	0.982	7.6	3.5	11.7	0.950	0.798	0.988	6.8	3.2	10.4	0.955	0.821	0.989	6.3	3.2	9.5
	Mean VBF	0.949	0.798	0.987	6.2	3.0	9.5	0.963	0.855	0.991	5.9	3.3	8.4	0.978	0.915	0.995	4.4	2.4	6.4	0.974	0.901	0.993	5.1	3.6	6.6
	PK HBF	0.877	0.501	0.969	14.6	8.8	20.4	0.907	0.647	0.977	13.5	7.0	20.0	0.934	0.750	0.983	11.9	6.0	17.9	0.946	0.795	0.986	10.5	5.4	15.7
	PK VBF	0.905	0.606	0.976	8.9	4.9	12.9	0.947	0.787	0.987	6.4	2.5	10.4	0.973	0.893	0.993	5.7	3.2	8.1	0.966	0.870	0.991	5.9	3.1	8.7
GCT	FFC GCT	0.334	-1.497	0.832	6.6	2.3	11.0	0.314	-1.866	0.831	6.6	2.5	10.8	0.440	-1.427	0.863	5.9	2.2	9.7	0.569	-0.784	0.894	5.1	1.8	8.4
	PFC GCT	0.788	-0.033	0.951	7.9	5.1	10.7	0.791	0.074	0.950	6.8	3.2	10.3	0.737	-0.013	0.934	7.5	3.9	11.2	0.774	0.052	0.945	6.8	3.4	10.2

Key: ADFM: Ankle dorsi-flexion moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFA: Ankle dorsi-flexion angle; KFA: Knee flexion angle; HFA: Hip flexion angle; FFC: Final foot contact; PFC: Penultimate foot contact; PK: Peak; GRF: Ground reaction force; KAA: Knee abduction angle; KAM: Knee abduction moment; Knee internal rotation moment; HBF: Horizontal braking force; VBF: Vertical braking force; GCT: Ground contact time; ICC: Intraclass correlation coefficient; CV%: Coefficient of variation. CI: Confidence interval; UB: Upper bound; LB: Lower bound.

Appendix 3.3 Outcome value comparisons between trial sizes (Chapter 4)

90˚ cut session 1 - outcome value comparisons between trial sizes

	Variable	*RMANOVA*	SET	TS3 vs TS5			TS3 vs TS8			TS3 vs TS10			TS5 vs TS8			TS5 vs TS10			TS8 vs TS10		
				p	*g*	*% dif*	*p*	*g*	*% dif*	*p*	*g*	*% dif*	*p*	*g*	*% dif*	*p*	*g*	*% dif*	*p*	*g*	*% dif*
FFC	PK ADFA (˚)	*0.062*	1	0.445	-0.02	-0.2	***0.037***	-0.13	-1.5	*0.093*	-0.15	-1.7	***0.017***	-0.12	-1.2	*0.074*	-0.14	-1.5	*0.575*	-0.03	-0.3
	PK KFA (˚)	0.321	1	1.000	0.06	0.6	1.000	0.15	1.5	1.000	0.16	1.5	0.965	0.09	0.9	1.000	0.10	0.9	1.000	0.00	0.0
	PK HFA (˚)	**<0.001**	6	0.599	0.09	1.8	0.090	0.22	4.3	0.055	0.25	5.0	0.109	0.12	2.6	0.106	0.16	3.3	1.000	0.03	0.7
	PK ADFM (Nm/kg)	0.166	1	1.000	-0.03	-0.6	1.000	-0.13	-2.9	0.862	-0.22	-5.1	1.000	-0.10	-2.3	0.848	-0.20	-4.5	0.721	-0.09	-2.1
	PK KFM (Nm/kg)	0.868	1	1.000	0.05	0.9	1.000	0.04	0.7	1.000	0.05	0.9	1.000	-0.01	-0.2	1.000	0.00	0.0	1.000	0.01	0.1
	PK HFM (Nm/kg)	0.508	1	1.000	0.07	2.3	1.000	0.08	2.7	1.000	0.05	1.5	1.000	0.01	0.4	1.000	-0.02	-0.8	1.000	-0.03	-1.2
	PK KAA (˚)	**0.002**	4	0.169	-0.08	-7.6	**0.045**	-0.13	-12.5	0.131	-0.12	-11.3	0.300	-0.05	-4.6	1.000	-0.04	-3.4	1.000	0.01	1.1
	PK KAM (Nm/kg)	0.456	1	1.000	-0.09	-3.2	1.000	-0.14	-4.8	1.000	-0.10	-3.8	1.000	-0.04	-1.5	1.000	-0.02	-0.6	1.000	0.03	0.9
	PK KIRM (Nm/kg)	0.884	1	0.957	0.01	0.4	0.724	-0.06	-2.3	0.665	-0.07	-2.6	.444	-0.07	-2.7	.494	-0.08	-2.9	0.919	-0.01	-0.2
	Mean HBF (N/BW)	0.204	4	1.000	-0.10	-1.8	0.370	-0.19	-3.1	0.351	-0.27	-4.2	1.000	-0.08	-1.3	0.762	-0.15	-2.4	0.707	-0.07	-1.0
	Mean VBF (N/BW)	0.558	1	1.000	-0.02	-0.2	1.000	-0.03	-0.5	1.000	-0.08	-1.2	1.000	-0.02	-0.3	1.000	-0.06	-0.9	1.000	-0.04	-0.6
	PK HBF (N/BW)	0.833	1	1.000	-0.03	-0.8	1.000	0.00	-0.1	1.000	-0.02	-0.6	1.000	0.02	0.7	1.000	0.01	0.2	1.000	-0.02	-0.5
	PK VBF (N/BW)	0.426	1	1.000	0.05	1.2	1.000	0.08	1.9	1.000	0.03	0.7	1.000	0.03	0.7	1.000	-0.02	-0.5	0.415	-0.05	-1.2
PFC	PK ADFA (˚)	0.885	1	1.000	0.01	0.1	1.000	-0.05	-0.5	1.000	-0.01	-0.1	1.000	-0.07	-0.7	1.000	-0.03	-0.3	1.000	0.04	0.4
	PK KFA (˚)	0.655	1	1.000	-0.02	-0.2	1.000	-0.05	-0.4	1.000	0.01	0.1	1.000	-0.02	-0.2	1.000	0.03	0.3	0.311	0.05	0.5
	PK HFA (˚)	*0.668*	1	*0.386*	-0.06	-0.7	*0.386*	-0.09	-1.0	*0.445*	-0.07	-0.9	*0.508*	-0.03	-0.3	*0.959*	-0.02	-0.2	*0.799*	0.01	0.1
	PK ADFM (Nm/kg)	0.955	1	1.000	0.00	0.0	1.000	-0.02	-0.9	1.000	-0.04	-1.4	1.000	-0.03	-0.9	1.000	-0.04	-1.4	1.000	-0.01	-0.5
	PK KFM (Nm/kg)	0.322	1	0.964	0.08	1.7	1.000	0.12	2.3	1.000	0.12	2.4	1.000	0.03	0.6	1.000	0.03	0.7	1.000	0.00	0.1
	PK HFM (Nm/kg)	*0.776*	1	*0.812*	-0.01	-0.5	*0.859*	-0.05	-2.1	*0.878*	0.01	0.4	*0.293*	-0.04	-1.6	*0.799*	0.02	0.9	*0.059*	0.06	2.4
	Mean HBF (N/BW)	0.699	2	1.000	0.01	0.4	1.000	0.00	0.1	1.000	-0.03	-0.8	1.000	-0.01	-0.3	1.000	-0.04	-1.2	1.000	-0.03	-0.9
	Mean VBF (N/BW)	0.718	1	1.000	0.03	0.8	1.000	0.05	1.3	1.000	0.03	0.6	1.000	0.02	0.5	1.000	-0.01	-0.2	0.829	-0.03	-0.7
	PK HBF (N/BW)	*0.782*	1	*0.386*	-0.03	-1.2	*0.508*	-0.05	-2.0	*0.285*	-0.08	-3.2	*0.575*	-0.02	-0.8	*0.333*	-0.05	-2.0	*0.508*	-0.03	-1.2
	PK VBF (N/BW)	0.559	3	1.000	-0.04	-0.9	1.000	-0.04	-1.0	1.000	-0.09	-2.4	1.000	0.00	-0.1	1.000	-0.06	-1.5	0.259	-0.06	-1.4
GCT	FFC GCT (s)	**0.049**	1	1.000	0.03	0.4	0.497	0.20	2.2	0.434	0.27	2.8	0.741	0.16	1.8	0.879	0.23	2.4	1.000	0.07	0.6
	PFC GCT (s)	0.454	2	1.000	0.01	0.1	1.000	0.06	0.9	1.000	0.15	2.1	1.000	0.06	0.8	0.965	0.14	1.9	0.064	0.08	1.2
		Average	1.6																		
		SD	1.3																		

Key: TS: Trial size; ADFM: Ankle dorsi-flexion moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFA: Ankle dorsi-flexion angle; KFA: Knee flexion angle; HFA: Hip flexion angle; FFC: Final foot contact; PFC: Penultimate foot contact; PK: Peak; GRF: Ground reaction force; KAA: Knee abduction angle; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; HBF: Horizontal braking force; VBF: Vertical braking force; GCT: Ground contact time; RMANOVA: Repeated measures ANOVA; SET: Sequential estimation technique; Bold denotes statistically significant difference. Italic denotes non parametric.

90˚ cut session 2 - outcome value comparisons between trial sizes

	Variable	*RMANOVA*	SET	TS3 vs TS5			TS3 vs TS8			TS3 vs TS10			TS5 vs TS8			TS5 vs TS10			TS8 vs TS10		
				p	*g*	*% dif*	*p*	*g*	*% dif*	*p*	*g*	*% dif*	*p*	*g*	*% dif*	*p*	*g*	*% dif*	*p*	*g*	*% dif*
FFC	PK ADFA (˚)	*0.923*	1	*0.575*	0.07	0.8	*0.575*	0.07	0.9	*0.575*	0.08	1.0	*0.878*	0.00	0.0	*0.799*	0.01	0.1	*0.541*	0.01	0.1
	PK KFA (˚)	0.404	1	1.000	0.05	0.7	1.000	0.06	0.8	1.000	0.11	1.5	1.000	0.01	0.1	1.000	0.06	0.9	0.396	0.05	0.7
	PK HFA (˚)	0.316	1	1.000	-0.01	-0.4	1.000	0.02	0.7	1.000	0.04	1.3	1.000	0.04	1.1	0.279	0.06	1.7	0.842	0.02	0.6
	PK ADFM (Nm/kg)	0.524	1	1.000	0.16	3.2	1.000	0.05	1.0	1.000	0.01	0.2	1.000	-0.11	-2.3	1.000	-0.15	-3.1	1.000	-0.04	-0.8
	PK KFM (Nm/kg)	0.331	1	1.000	0.08	1.2	1.000	0.05	0.7	1.000	0.11	1.5	1.000	-0.03	-0.5	1.000	0.02	0.4	0.105	0.06	0.8
	PK HFM (Nm/kg)	*0.231*	1	*0.184*	-0.09	-2.7	*0.332*	-0.09	-2.8	*0.102*	-0.14	-4.2	*0.726*	-0.01	-0.2	*0.286*	-0.05	-1.5	*0.092*	-0.05	-1.3
	PK KAA (˚)	*0.052*	4	*0.093*	-0.07	-7.7	***0.037***	-0.11	-11.1	***0.028***	-0.13	-13.2	*0.139*	-0.03	-3.2	*0.074*	-0.05	-5.2	*0.314*	-0.02	-2.0
	PK KAM (Nm/kg)	*0.110*	1	*0.306*	-0.06	-3.0	*0.779*	-0.03	-1.5	*0.574*	-0.07	-3.5	*0.084*	0.03	1.4	*0.767*	-0.01	-0.5	*0.065*	-0.04	-1.9
	PK KIRM (Nm/kg)	0.108	1	0.165	-0.25	-10.0	0.129	-0.33	-13.6	0.070	-0.37	-14.4	0.192	-0.08	-3.3	0.087	-0.10	-4.1	0.692	-0.02	-0.7
	Mean HBF (N/BW)	0.699	2	1.000	-0.01	-0.2	1.000	-0.02	-0.5	1.000	-0.06	-1.3	1.000	-0.01	-0.3	1.000	-0.05	-1.1	0.285	-0.04	-0.9
	Mean VBF (N/BW)	0.837	1	1.000	-0.01	-0.1	1.000	0.01	0.3	1.000	0.00	-0.1	1.000	0.02	0.4	1.000	0.00	0.0	1.000	-0.02	-0.4
	PK HBF (N/BW)	0.338	2	1.000	-0.03	-1.2	1.000	-0.04	-1.4	1.000	-0.07	-2.3	1.000	-0.01	-0.2	1.000	-0.03	-1.1	0.513	-0.03	-0.9
	PK VBF (N/BW)	0.952	1	1.000	0.00	0.1	1.000	0.02	0.7	1.000	0.01	0.3	1.000	0.02	0.6	1.000	0.01	0.2	1.000	-0.01	-0.4
PFC	PK ADFA (˚)	0.301	1	1.000	0.06	0.7	1.000	0.11	1.4	1.000	0.16	2.0	1.000	0.05	0.7	1.000	0.11	1.3	1.000	0.06	0.7
	PK KFA (˚)	0.795	1	1.000	-0.03	-0.4	1.000	-0.02	-0.2	1.000	-0.01	-0.2	1.000	0.02	0.2	1.000	0.02	0.2	1.000	0.00	0.0
	PK HFA (˚)	0.412	1	1.000	-0.03	-0.4	1.000	-0.06	-0.8	1.000	-0.07	-0.9	1.000	-0.03	-0.4	1.000	-0.04	-0.5	1.000	-0.01	-0.1
	PK ADFM (Nm/kg)	0.519	1	1.000	-0.03	-1.3	1.000	-0.01	-0.5	1.000	0.06	2.3	1.000	0.02	0.8	1.000	0.10	3.6	0.161	0.08	2.8
	PK KFM (Nm/kg)	*0.743*	1	*0.415*	-0.05	-1.0	*0.646*	-0.03	-0.7	*0.575*	-0.06	-1.1	*0.759*	0.01	0.3	*0.959*	-0.01	-0.1	*0.512*	-0.02	-0.4
	PK HFM (Nm/kg)	0.816	1	1.000	0.00	-0.1	1.000	0.01	0.8	1.000	0.04	2.1	1.000	0.02	0.8	1.000	0.05	2.2	1.000	0.03	1.4
	Mean HBF (N/BW)	0.537	2	1.000	0.04	0.9	1.000	0.01	0.2	1.000	-0.02	-0.4	1.000	-0.03	-0.7	1.000	-0.05	-1.4	1.000	-0.02	-0.6
	Mean VBF (N/BW)	0.659	1	1.000	0.03	0.7	1.000	0.03	0.7	1.000	0.04	1.1	1.000	0.00	-0.1	1.000	0.01	0.3	1.000	0.02	0.4
	PK HBF (N/BW)	0.422	1	1.000	0.03	1.2	1.000	0.00	0.1	1.000	-0.01	-0.6	1.000	-0.03	-1.1	0.978	-0.05	-1.8	1.000	-0.02	-0.7
	PK VBF (N/BW)	0.656	1	1.000	0.05	1.3	1.000	0.05	1.5	1.000	0.05	1.3	1.000	0.00	0.1	1.000	0.00	0.0	1.000	-0.01	-0.2
GCT	FFC GCT (s)	0.544	1	1.000	-0.06	-0.6	1.000	0.01	0.2	1.000	0.06	0.7	1.000	0.07	0.8	0.601	0.12	1.3	0.578	0.04	0.5
	PFC GCT (s)	0.321	1	1.000	-0.06	-0.9	1.000	0.03	0.4	1.000	0.07	1.0	0.954	0.09	1.2	0.516	0.14	1.9	1.000	0.05	0.6
		Average	1.3																		
		SD	0.7																		

Key: TS: Trial size; ADFM: Ankle dorsi-flexion moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFA: Ankle dorsi-flexion angle; KFA: Knee flexion angle; HFA: Hip flexion angle; FFC: Final foot contact; PFC: Penultimate foot contact; PK: Peak; GRF: Ground reaction force; KAA: Knee abduction angle; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; HBF: Horizontal braking force; VBF: Vertical braking force; GCT: Ground contact time; RMANOVA: Repeated measures ANOVA; SET: Sequential estimation technique; Bold denotes statistically significant difference. Italic denotes non parametric.

90˚ cut between-session - outcome value comparisons between trial sizes

	Variable	*RMANOVA*	TS3 vs TS5			TS3 vs TS8			TS3 vs TS10			TS5 vs TS8			TS5 vs TS10			TS8 vs TS10		
			p	*g*	*% dif*	*p*	*g*	*% dif*	*p*	*g*	*% dif*	*p*	*g*	*% dif*	*p*	*g*	*% dif*	*p*	*g*	*% dif*
	PK ADFA (˚)	*0.954*	*0.386*	0.03	0.3	*0.515*	-0.03	-0.3	*0.508*	-0.03	-0.4	*0.114*	-0.05	-0.6	*0.114*	-0.06	-0.7	*0.646*	-0.01	-0.1
	PK KFA (˚)	0.210	1.000	0.06	0.6	1.000	0.11	1.1	0.882	0.14	1.5	1.000	0.05	0.5	1.000	0.08	0.9	0.480	0.04	0.4
	PK HFA (˚)	**<0.001**	1.000	0.03	0.7	0.123	0.11	2.6	0.035	0.14	3.2	0.133	0.08	1.9	0.059	0.11	2.5	0.578	0.03	0.6
	PK ADFM (Nm/kg)	0.204	1.000	0.08	1.3	1.000	-0.05	-0.9	1.000	-0.13	-2.4	0.813	-0.12	-2.3	0.404	-0.20	-3.8	0.313	-0.07	-1.4
	PK KFM (Nm/kg)	0.780	1.000	0.07	1.0	1.000	0.05	0.7	1.000	0.08	1.2	1.000	-0.02	-0.3	1.000	0.01	0.2	0.890	0.03	0.5
	PK HFM (Nm/kg)	*0.303*	*0.859*	0.00	0.0	*0.953*	0.00	0.1	*0.374*	-0.04	-1.2	*0.833*	0.00	0.1	*0.167*	-0.04	-1.1	*0.058*	-0.04	-1.3
FFC	PK KAA (˚)	**0.007**	**0.017**	-0.08	-7.6	0.007	-0.12	-11.9	0.005	-0.13	-12.1	0.009	-0.04	-4.0	0.059	-0.05	-4.2	0.721	0.00	-0.2
	PK KAM (Nm/kg)	*0.396*	*0.152*	-0.08	-3.1	*0.202*	-0.08	-3.3	*0.241*	-0.09	-3.6	*0.512*	0.00	-0.2	*0.539*	-0.01	-0.5	*0.809*	-0.01	-0.3
	PK KIRM (Nm/kg)	0.294	0.417	-0.13	-4.3	0.313	-0.20	-7.4	0.202	-0.22	-7.9	0.277	-0.08	-3.0	0.212	-0.10	-3.5	0.722	-0.01	-0.5
	Mean HBF (N/BW)	0.310	1.000	-0.05	-1.0	1.000	-0.09	-1.7	1.000	-0.14	-2.7	1.000	-0.04	-0.8	1.000	-0.10	-1.7	0.256	-0.06	-0.9
	Mean VBF (N/BW)	0.709	1.000	-0.01	-0.2	1.000	-0.01	-0.1	1.000	-0.03	-0.6	1.000	0.00	0.1	1.000	-0.02	-0.4	0.399	-0.03	-0.5
	PK HBF (N/BW)	0.586	1.000	-0.03	-1.0	1.000	-0.02	-0.7	1.000	-0.05	-1.5	1.000	0.01	0.3	1.000	-0.01	-0.5	0.256	-0.02	-0.7
	PK VBF (N/BW)	0.645	1.000	0.03	0.7	1.000	0.05	1.3	1.000	0.02	0.5	1.000	0.02	0.6	1.000	-0.01	-0.2	0.687	-0.03	-0.8
	PK ADFA (˚)	0.583	1.000	0.04	0.4	1.000	0.04	0.4	1.000	0.09	0.9	1.000	0.00	0.0	1.000	0.05	0.5	1.000	0.05	0.5
	PK KFA (˚)	0.664	1.000	-0.03	-0.3	1.000	-0.03	-0.3	1.000	0.00	0.0	1.000	0.00	0.0	1.000	0.02	0.3	0.640	0.03	0.3
	PK HFA (˚)	0.102	1.000	-0.05	-0.6	0.355	-0.08	-0.9	0.638	-0.08	-0.9	1.000	-0.03	-0.3	1.000	-0.03	-0.3	1.000	0.00	0.0
	PK ADFM (Nm/kg)	0.846	1.000	-0.02	-0.6	1.000	-0.02	-0.7	1.000	0.02	0.5	1.000	0.00	0.0	1.000	0.03	1.2	1.000	0.03	1.2
PFC	PK KFM (Nm/kg)	*0.954*	*0.609*	0.02	0.4	*0.678*	0.05	0.9	*0.878*	0.04	0.7	*0.646*	0.02	0.5	*0.799*	0.01	0.3	*0.720*	-0.01	-0.2
	PK HFM (Nm/kg)	0.739	1.000	-0.01	-0.3	1.000	-0.01	-0.6	1.000	0.03	1.3	1.000	-0.01	-0.3	1.000	0.04	1.6	0.455	0.05	1.9
	Mean HBF (N/BW)	0.566	1.000	0.03	0.7	1.000	0.01	0.1	1.000	-0.02	-0.6	1.000	-0.02	-0.5	1.000	-0.05	-1.3	1.000	-0.03	-0.7
	Mean VBF (N/BW)	0.546	0.564	0.03	0.8	1.000	0.04	1.0	1.000	0.04	0.8	1.000	0.01	0.2	1.000	0.00	0.1	1.000	-0.01	-0.1
	PK HBF (N/BW)	*0.668*	*0.799*	0.00	0.0	*0.646*	-0.03	-0.9	*0.445*	-0.05	-1.9	*0.285*	-0.03	-0.9	*0.333*	-0.05	-1.9	*0.386*	-0.03	-1.0
	PK VBF (N/BW)	0.653	1.000	0.01	0.2	1.000	0.01	0.2	1.000	-0.02	-0.6	1.000	0.00	0.0	1.000	-0.03	-0.8	0.427	-0.03	-0.8
GCT	FFC GCT (s)	0.164	1.000	-0.01	-0.1	1.000	0.14	1.2	1.000	0.20	1.7	0.672	0.15	1.3	0.507	0.22	1.9	0.598	0.07	0.6
	PFC GCT (s)	0.126	1.000	-0.03	-0.3	1.000	0.05	0.7	1.000	0.12	1.6	1.000	0.08	1.0	0.570	0.15	1.9	0.230	0.07	0.9

Key: TS: Trial size; ADFM: Ankle dorsi-flexion moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFA: Ankle dorsi-flexion angle; KFA: Knee flexion angle; HFA: Hip flexion angle; FFC: Final foot contact; PFC: Penultimate foot contact; PK: Peak; GRF: Ground reaction force; KAA: Knee abduction angle; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; HBF: Horizontal braking force; VBF: Vertical braking force; GCT: Ground contact time; RMANOVA: Repeated measures ANOVA; Bold denotes statistically significant difference. Italic denotes non parametric.

Appendix 4.1. Correlations (Chapter 5)

Short cut correlations

	Description	Variable	Mean	SD	*Completion time*	FFC GCT	Exit velocity	*pk KAM*	*pk KIRM*
					ρ	***r or ρ***	***r or ρ***	***ρ***	***ρ***
		Completion time (s)	2.101	0.164	-	_.361**_	_-.621**_	_-.345**_	_.340**_
Sag joint moment		FFC pk HFM (Nm/kg)	-2.90	0.83	_.333**_	0.109	-0.077	_-0.250_	_.411**_
		FFC pk KFM (Nm/kg)	3.41	0.66	_-.316*_	-.495**	.557**	_.338**_	_-0.199_
		FFC pk ADFM (Nm/kg)	-1.73	0.55	_0.006_	_.293*_	_-0.212_	_-.264*_	_0.227_
Sag joint angles	(+ flexion/ -extension)	FFC HFA (°) - pk	51.7	11.0	_0.165_	.554**	-.309*	_-.259*_	_0.247_
		FFC HFA (°) - IC	47.1	8.7	_0.072_	.330**	-0.151	_-0.105_	_0.051_
		FFC HFA (°) - ROM	4.6	4.5	_.274*_	_.654**_	_-.431**_	_-.526**_	_.490**_
		FFC KFA (°) - pk	64.2	7.7	_0.073_	.621**	-0.150	_-0.214_	_0.201_
		FFC KFA (°) - IC	24.2	5.3	_-0.079_	-0.061	0.144	_0.079_	_-0.111_
		FFC KFA (°) - ROM	40.0	7.5	_0.153_	.684**	-.257*	_-.290*_	_.358**_
		FFC ADFA (°) - pk	77.9	8.3	_-0.007_	0.048	0.131	_0.054_	_0.130_
		FFC ADFA (°) - IC	56.3	9.5	_0.099_	.477**	-0.144	_-0.222_	_.442**_
		FFC ADFA (°) - ROM	21.6	10.9	_-0.090_	-.378**	0.224	_.311*_	_-.340**_
Inj-risk parameters	(+abduction/ - adduction)	pk KAM (Nm/kg)	1.00	0.34	_-.345**_	_-.390**_	_.396**_	-	_-.522**_
	(-abduction/ + adduction)	pk KAA (°)	-11.1	5.9	_0.136_	.272*	-0.226	_-.445**_	_.308*_
		KAA (°) - IC	2.8	4.5	_-0.018_	.274*	-0.144	_-.326*_	_0.236_
	(-internal/ + external)	pk KIRM (Nm/kg)	-0.90	0.44	_.340**_	_.495**_	_-.334**_	_-.522**_	-
		pk KRA (°)	-4.3	8.1	_0.043_	-.278*	0.133	_0.082_	_-0.099_
		KRA (°) - IC	-2.8	8.7	_0.092_	-.299*	0.143	_0.162_	_-0.139_
Trunk	(+forward/ - backward)	PFC Sag trunk inclination angle - IC (°)	9.5	4.9	_-.441**_	_-.399**_	_0.231_	_0.241_	_-.271*_
		FFC Sag trunk inclination angle - IC (°)	11.9	7.2	_-.303*_	_-.317*_	_0.160_	_0.124_	_-0.089_
	(-lateral/ +medial)	Lateral trunk flexion (°) - IC	-19.4	7.5	_0.042_	-0.107	0.113	_-0.022_	_0.128_
		Lateral trunk flexion (°) - pk	-29.8	9.0	_-0.077_	-.531**	.277*	_0.183_	_-0.093_
		Lateral Trunk flexion (°) - ROM	10.3	5.6	_0.207_	.711**	-.293*	_-.364**_	_.404**_
Hip, pelvis, foot, technique	(-internal/ + external	Hip rotation angle (°) - IC	13.5	9.0	_0.076_	-0.018	0.028	_0.036_	_0.064_
		Hip rotation angle (°) - pk	12.2	9.0	_0.064_	-0.020	0.013	_0.001_	_0.121_
	(-abduction/ + adduction)	Hip abduction angle (°) - IC	-22.8	5.5	_-0.035_	0.013	-0.058	_-0.049_	_-0.033_
	(+ towards intended COD/ - away	Pelvic rotation (°) - IC	29.2	9.2	_-0.176_	-0.232	0.227	_0.144_	_-0.032_
	(+ internal/ - external)	IFPA (°) - IC	15.1	10.3	_-0.242_	-.295*	.260*	_0.187_	_-0.233_
		Lateral foot plant distance (m)	-0.314	0.051	_0.117_	.579**	-.257*	_-.280*_	_.365**_
Braking GRF		PFC HBF - pk (BW)	-1.43	0.38	_0.236_	_-0.028_	_0.020_	_0.048_	_0.236_
		PFC HBF - mean (BW)	-0.54	0.10	_.449**_	0.195	-0.137	_-0.065_	_.302*_
		FFC VBF - pk (BW)	2.56	0.59	_-0.236_	_-.493**_	_.257*_	_.333**_	_-.468**_
		FFC VBF - mean (BW)	1.60	0.22	_-0.231_	-.716**	.397**	_.389**_	_-.450**_
		FFC HBF - pk (BW)	-1.38	0.34	_0.195_	0.246	-0.130	_-.323*_	_.361**_
		FFC HBF - mean (BW)	-0.82	0.16	_.258*_	.609**	-.358**	_-.435**_	_.507**_
		PFC RBF - pk (BW)	2.88	0.58	_-0.224_	_-0.038_	_-0.079_	_0.009_	_-0.196_
		PFC RBF - mean (BW)	1.09	0.15	_-0.121_	-0.202	0.013	_-0.112_	_-0.091_
		FFC RBF - pk (BW)	3.00	0.64	_-.260*_	_-.548**_	_.302*_	_.360**_	_-.495**_
		FFC RBF - mean (BW)	1.94	0.29	_-.273*_	-.743**	.436**	_.423**_	_-.501**_
Propulsive GRF		FFC VPF - pk (BW)	1.82	0.25	_-.304*_	_-.811**_	_.538**_	_0.204_	_-0.218_
		FFC VPF - mean (BW)	1.21	0.11	_-.268*_	-.508**	.512**	_0.182_	_0.009_
		FFC HPF - pk (BW)	-0.76	0.20	_.358**_	.789**	-.488**	_-.385**_	_.395**_
		FFC HPF - mean (BW)	-0.35	0.10	_.396**_	_.611**_	_-.462**_	_-.306*_	_0.235_
		FFC MLPF - pk (BW)	1.09	0.18	_-.499**_	-.752**	.687**	_0.220_	_-.361**_
		FFC MLPF - mean (BW)	0.76	0.11	_-.609**_	-.610**	.726**	_0.250_	_-.285*_
		RPF - pk (BW)	2.22	0.34	_-.378**_	_-.853**_	_.585**_	_.255*_	_-.314*_
		RPF- mean (BW)	1.49	0.16	_-.424**_	-.603**	.611**	_0.216_	_-0.120_
		pk HBF ratio	1.01	0.31	_0.058_	_-.374**_	_0.114_	_.350**_	_-0.105_
		mean HBF ratio	1.56	0.46	_0.039_	_-.277*_	_0.184_	_.341**_	_-0.172_
GCT		PFC GCT (s)	0.201	0.040	_0.228_	_.639**_	_-.403**_	_-.263*_	_.384**_
		FFC GCT (s)	0.319	0.058	_.361**_	-	_-.547**_	_-.390**_	_.495**_
Velocity profile		Approach velocity (m/s)	4.30	0.34	_-.413**_	-0.250	.456**	_.324*_	_-.476**_
		FFC touch-down velocity (m/s)	3.27	0.34	_-.345**_	-.597**	.570**	_.527**_	_-.588**_
		EXIT velocity (m/s)	3.24	0.25	_-.621**_	-.547**	-	_.396**_	_-.334**_
		Δ PFC velocity (m/s)	-1.03	0.23	_0.049_	-.510**	0.166	_.333**_	_-0.153_
		Δ FFC velocity (m/s)	-0.89	0.27	_-0.207_	-.345**	.254*	_-0.079_	_0.160_

Moderate correlation (0.30-0.49) | **Large correlation (0.50-0.69)** | **Very large correlation (0.70-0.89)**

Key: pk: peak; GCT: Ground contact time; PFC: Penultimate foot contact; FFC: Final foot contact; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFM: Ankle-dorsi flexion moment; HFA: Hip flexion angle; KFA: Knee flexion angle; ADFA: Ankle dorsi-flexion angle; RPF: Resultant propulsive force; RBF: Resultant braking force; VPF: Vertical propulsive force: VBF: Vertical braking force; HPF: Horizontal propulsive force; HBF: Horizontal braking force; MLPF: Medio-lateral propulsive force; ROM: Range of motion; IC: Initial contact; BW: Body weight; IFPA: Initial foot progression angle; Italic denotes non-parametric.

Long cut correlations

Description		Variable	Mean	SD	Completion Time	FFC GCT	Exit velocity	pk KAM	pk KRM
					r or ρ	ρ	r or ρ	r or ρ	ρ
		Completion time (s)	1.759	0.135	-	.581**	-.733**	-.412**	.539**
Sag joint moment		FFC pk HFM (Nm/kg)	-2.72	0.72	0.226	-0.054	-0.030	-0.152	.255*
		FFC pk KFM (Nm/kg)	3.40	0.68	-.295*	-.447**	.482**	.317*	-0.151
		FFC pk ADFM (Nm/kg)	-1.84	0.49	.291*	.430**	-.282*	-0.180	0.196
Sag joint angles	(+ flexion/ -extension)	FFC HFA (˚) - pk	47.5	11.2	.332**	.603**	-.385**	-.307*	0.171
		FFC HFA (˚) - IC	43.1	8.6	0.144	.450**	-0.198	-0.190	0.032
		FFC HFA (˚) - ROM	4.4	4.6	.406**	.658**	-.470**	-.311*	.281*
		FFC KFA (˚) - pk	62.5	7.2	0.222	.616**	-0.152	-0.172	0.149
		FFC KFA (˚) - IC	23.1	4.6	-0.065	0.043	0.179	0.059	-0.158
		FFC KFA (˚) - ROM	39.4	6.6	.286*	.670**	-.289*	-0.227	0.214
		FFC ADFA (˚) - pk	78.3	8.2	-0.045	-0.017	0.160	0.013	0.085
		FFC ADFA (˚) - IC	55.7	10.5	0.231	.441**	-0.047	-0.123	.321*
		FFC ADFA (˚) - ROM	22.5	11.2	-0.249	-.451**	0.160	0.125	-.291*
Inj-risk parameters	(+abduction/ - adduction)	pk KAM (Nm/kg)	1.11	0.39	-.412**	-.308*	.350**	-	-.557**
	(-abduction/ + adduction)	pk KAA (°)	-11.6	6.5	0.060	-0.001	-0.034	-.468**	0.252
		KAA (°) - IC	2.5	4.9	0.040	0.171	-0.065	-.301*	0.218
	(-internal/ + external)	pk KIRM (Nm/kg)	-0.94	0.44	.539**	.396**	-.367**	-.557**	-
		pk KRA (°)	-3.9	9.1	-0.069	-0.249	0.137	0.043	-0.106
		KRA (°) - IC	-2.5	10.0	-0.044	-0.201	0.101	0.093	-0.149
Trunk	(+forward/ - backward)	PFC Sag trunk inclination angle - IC (°)	10.0	4.8	-.423**	-.295*	.295*	-0.011	-0.156
		FFC Sag trunk inclination angle - IC (°)	12.7	6.7	-0.146	-0.180	0.092	-0.092	0.053
	(-lateral/ +medial)	Lateral trunk flexion (˚) - IC	-15.2	7.9	-0.131	-.330**	0.036	0.062	0.000
		Lateral trunk flexion (˚) - pk	-26.6	9.9	-0.252	-.623**	0.233	0.230	-0.047
		Lateral Trunk flexion (˚) - ROM	11.4	5.8	0.188	.595**	-.291*	-.256*	0.170
Hip, pelvis, foot, technique	(-internal/ + external	Hip rotation angle (°) - IC	15.0	10.5	0.177	0.165	-0.105	0.093	0.148
		Hip rotation angle (°) - pk	13.4	9.8	0.180	0.113	-0.082	0.057	0.195
	(-abduction/ + adduction)	Hip abduction angle (°) - IC	-24.3	6.3	0.082	-0.088	-0.015	-0.151	-0.002
	(+ towards intended COD/- away)	Pelvic rotation (°) - IC	33.8	9.3	-.318*	-.264*	0.209	0.175	-0.018
	(+ internal/ - external)	IFPA (°) - IC	17.4	10.3	-.411**	-.383**	.283*	0.158	-0.139
		Lateral foot plant distance (m)	-0.308	0.051	0.218	.626**	-0.227	-0.239	0.176
Braking GRF		PFC HBF - pk (BW)	-1.66	0.40	.260*	0.053	-0.172	-0.044	.254*
		PFC HBF - mean (BW)	-0.60	0.12	.551**	.343**	-.430**	-0.146	.452**
		FFC VBF - pk (BW)	2.56	0.52	-.271*	-.411**	0.189	.314*	-.475**
		FFC VBF - mean (BW)	1.57	0.18	-.389**	-.598**	.456**	.497**	-.468**
		FFC HBF - pk (BW)	-1.44	0.33	.270*	0.170	-0.158	-.279*	.369**
		FFC HBF - mean (BW)	-0.85	0.16	.535**	.560**	-.358**	-.434**	.433**
		PFC RBF - pk (BW)	3.06	0.58	-.259*	-0.141	0.226	0.157	-.303*
		PFC RBF - mean (BW)	1.15	0.18	-.287*	-.326*	.321*	0.057	-0.248
		FFC RBF - pk (BW)	3.03	0.59	-.280*	-.411**	0.187	.306*	-.458**
		FFC RBF - mean (BW)	1.93	0.25	-.484**	-.643**	.472**	.488**	-.505**
Propulsive GRF		FFC VPF - pk (BW)	1.82	0.27	-.449**	-.846**	.540**	0.211	-0.177
		FFC VPF - mean (BW)	1.20	0.11	-.460**	-.610**	.499**	0.115	-0.021
		FFC HPF - pk (BW)	-0.81	0.22	.608**	.818**	-.500**	-.334**	.269*
		FFC HPF - mean (BW)	-0.37	0.11	.588**	.693**	-.336**	-0.227	0.157
		FFC MLPF - pk (BW)	1.09	0.20	-.627**	-.823**	.651**	0.243	-.277*
		FFC MLPF - mean (BW)	0.75	0.11	-.641**	-.704**	.638**	0.159	-0.221
		RPF - pk (BW)	2.24	0.38	-.530**	-.878**	.568**	0.247	-0.224
		RPF- mean (BW)	1.48	0.17	-.579**	-.695**	.549**	0.166	-0.121
		pk HBF ratio	0.90	0.22	0.041	-0.053	-0.053	0.213	-0.043
		mean HBF ratio	1.47	0.31	0.034	-0.173	-0.003	0.197	-0.006
GCT		PFC GCT (s)	0.202	0.041	.551**	.711**	-.484**	-.370**	.436**
		FFC GCT (s)	0.307	0.058	.592**	-	-.582**	-.306*	.394**
Velocity profile		Approach velocity (m/s)	4.58	0.41	-.660**	-.306*	.533**	.384**	-.534**
		FFC touch-down velocity (m/s)	3.45	0.37	-.752**	-.558**	.725**	.491**	-.551**
		EXIT velocity (m/s)	3.30	0.30	-.733**	-.569**	-	.350**	-.367**
		Δ PFC velocity (m/s)	-1.13	0.19	-0.045	-.425**	0.165	0.124	0.061
		Δ FFC velocity (m/s)	-0.90	0.22	-0.169	-.464**	.365**	-0.027	0.118
		Approach time (s)	1.980	0.160	.632**	.325*	-.496**	-0.185	.289*

Moderate correlation (0.30-0.49) | **Large correlation (0.50-0.69)** | **Very large correlation (0.70-0.89)**

Key: pk: peak; GCT: Ground contact time; PFC: Penultimate foot contact; FFC: Final foot contact; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFM: Ankle-dorsi flexion moment; HFA: Hip flexion angle; KFA: Knee flexion angle; ADFA: Ankle dorsi-flexion angle; RPF: Resultant propulsive force; RBF: Resultant braking force; VPF: Vertical propulsive force: VBF: Vertical braking force; HPF: Horizontal propulsive force; HBF: Horizontal braking force; MLPF: Medio-lateral propulsive force; ROM: Range of motion; IC: Initial contact; BW: Body weight; IFPA: Initial foot progression angle. Italic denotes non-parametric.

Appendix 4.2 Biomechanical determinants of knee joint loads normalised for approach velocity (Chapter 5)

Peak KAMs Normalised to approach velocity: Short cut

A moderate and significant relationship was observed between peak KAMs and peak KIRMs (ρ = -0.557, p < 0.001) normalised to approach velocity. Greater peak KAMs normalised to approach velocity were associated with smaller reductions in velocity over PFC (ρ = 0.452, p < 0.001), lower hip flexion ROM (ρ = -0.489, p < 0.001), and greater peak KAA (ρ = -0.465, p < 0.001).

Peak KIRMs Normalised to approach velocity: Short cut

A large and significant relationship was observed between peak KIRMs normalised to approach velocity and FFC touch-down velocity (ρ = -0.511, p < 0.001). Greater peak KIRMs normalised to approach velocity were associated with lower hip flexion ROM (ρ = 0.487, p < 0.001), lower knee flexion ROM (r = 0.404, p = 0.001), smaller ADFA IC (r = 0.453, p < 0.001), and shorter FFC GCTs (r = -0.495, p < 0.001). Additionally, greater braking variables including FFC peak VBF (ρ = -0.428, p = 0.001), FFC mean VBF (r = -0.418, p = 0.001), FFC mean HBF (r = 0.462, p < 0.001), FFC peak (r = -0.456, p < 0.001) and mean (r = -0.466, p < 0.001) RBF were associated with greater peak KIRMs normalised to approach velocity and these were classed as moderate.

Peak KAMs Normalised to approach velocity: Long cut

A large and significant relationship was observed between peak KAMs normalised to approach velocity and peak KAA (r = -0.537, p < 0.001). Greater peak KAMs normalised to approach velocity were moderately associated with greater FFC mean VBFs (r = 0.402, p = 0.001) and greater normalised KIRMs (r = -0.440, p < 0.001).

Peak KIRMs Normalised to approach velocity: Long cut

Greater peak KIRMs normalised to approach velocity were moderately associated with shorter completion times (ρ = -0.468, p < 0.001), greater FFC mean BRF (ρ = -0.451, p < 0.001), greater peak KAMs normalised to approach velocity (ρ = -0.440, p < 0.001), greater FFC touch down velocity (ρ = -0.439, p < 0.001), and greater FFC peak (ρ = -0.415, p = 0.001) and mean VBF (ρ = -0.420, p = 0.001).

Appendix 4.3 Fast versus slow and low and high KAM comparisons – short cut and long cut (Chapter 5)

Short cut

	Completion time							Exit velocity							peak KAM						
	Fast (n = 20)		Slow (n = 20)		*p*	*g*	95% CI	Fast (n = 20)		Slow (n = 20)		*p*	*g*	95% CI	Low (n = 20)		High (n = 20)		*p*	*g*	95% CI
Variable	Mean	SD	Mean	SD				Mean	SD	Mean	SD				Mean	SD	Mean	SD			
Completion time (s)	1.950	0.064	2.268	0.161	<0.001	-2.55	0.83	2.038	0.199	2.205	0.136	*<0.001*	-0.96	0.63	2.144	0.143	2.048	0.211	*0.210*	0.52	0.65
FFC pk HFM (Nm/kg)	-3.24	0.83	-2.69	0.66	0.027	-0.71	0.64	-2.88	0.72	-3.00	0.94	0.669	0.13	0.63	-2.79	0.75	-3.15	1.05	0.223	0.38	0.62
FFC pk KFM (Nm/kg)	3.63	0.65	3.29	0.76	0.280	0.47	0.63	3.90	0.68	3.02	0.36	*<0.001*	1.60	0.64	3.19	0.53	3.65	0.60	0.014	-0.80	0.71
FFC pk ADFM (Nm/kg)	-1.78	0.48	-1.76	0.56	0.885	-0.04	0.62	-1.93	0.66	-1.55	0.44	0.039	-0.66	0.63	-1.63	0.59	-1.88	0.47	0.140	0.47	0.64
FFC HFA (°) - pk	49.5	8.9	52.6	10.9	0.331	-0.30	0.62	48.7	8.9	56.1	7.9	0.008	-0.86	0.64	55.0	10.6	47.1	11.0	*0.096*	0.72	0.65
FFC HFA (°) - IC	46.7	8.0	47.2	7.4	0.845	-0.06	0.62	46.3	8.9	49.0	4.5	0.241	-0.37	0.63	48.2	8.1	44.9	9.5	0.237	0.37	0.62
FFC HFA (°) - ROM	2.8	3.2	5.4	5.8	0.183	-0.55	0.63	2.4	2.1	7.2	5.9	*0.007*	-1.06	0.67	6.8	5.0	2.2	2.6	*<0.001*	1.13	0.66
FFC KFA (°) - pk	63.5	7.1	64.5	7.7	0.687	-0.12	0.62	63.6	8.5	64.8	7.3	0.628	-0.15	0.63	66.0	7.2	62.9	7.3	0.195	0.41	0.62
FFC KFA (°) - IC	24.5	5.2	23.9	5.3	0.726	0.11	0.62	26.1	5.3	23.4	5.4	0.109	0.51	0.62	23.8	4.5	25.5	5.1	0.274	-0.34	0.63
FFC KFA (°) - ROM	39.0	7.7	40.6	8.4	0.549	-0.19	0.62	37.4	7.5	41.4	7.7	0.157	-0.52	0.64	42.2	6.9	37.4	7.2	0.040	0.66	0.63
FFC ADFA (°) - pk	78.5	9.3	77.7	7.6	0.772	0.09	0.62	80.3	9.3	75.7	8.3	0.110	0.51	0.62	77.3	8.8	79.0	6.9	0.478	-0.22	0.63
FFC ADFA (°) - IC	55.8	8.6	57.1	9.5	0.637	-0.15	0.62	54.9	9.4	58.1	9.2	0.383	-0.33	0.63	58.2	9.9	53.0	8.5	0.087	0.55	0.62
FFC ADFA (°) - ROM	22.8	9.7	20.6	12.2	0.783	0.19	0.62	25.4	9.0	17.6	11.5	*0.020*	0.73	0.64	19.1	11.3	26.0	9.3	0.042	-0.65	0.64
pk KAM (Nm/kg)	1.12	0.32	0.95	0.37	0.131	0.48	0.63	1.07	0.32	0.84	0.29	0.022	0.74	1.16	0.65	0.10	1.41	0.21	*<0.001*	-4.49	0.64
pk KAA (°)	-12.4	5.6	-11.2	6.3	0.529	-0.20	0.62	-12.8	5.1	-8.9	5.9	0.029	-0.70	0.66	-8.5	4.4	-14.1	6.2	0.002	1.01	0.64
KAA (°) - IC	3.1	3.7	2.3	5.2	0.545	0.19	0.62	1.1	4.9	4.0	4.3	0.055	-0.62	0.64	4.1	3.8	1.2	5.0	0.044	0.64	0.63
pk KIRM (Nm/kg)	-1.01	0.33	-0.79	0.44	0.024	-0.57	0.63	-1.03	0.51	-0.74	0.36	*0.046*	-0.65	0.68	-0.68	0.32	-1.15	0.41	<0.001	1.24	0.64
pk KRA (°)	-5.5	6.9	-4.4	10.4	0.706	-0.12	0.62	-2.6	7.2	-5.0	9.3	0.370	0.28	0.62	-4.6	6.6	-4.1	8.3	0.840	-0.07	0.62
KRA (°) - IC	-4.7	7.3	-2.8	10.6	0.509	-0.21	0.62	-0.8	8.8	-4.0	9.5	0.270	0.35	0.62	-3.8	7.1	-1.7	9.0	0.431	-0.25	0.62
PFC Sag trunk inclination angle - IC (°)	11.9	5.1	7.8	5.0	0.002	0.79	0.64	10.9	5.4	8.5	4.2	0.129	0.48	0.63	8.8	4.8	10.9	5.0	0.190	-0.41	0.63
FFC Sag trunk inclination angle - IC (°)	14.7	7.4	9.6	7.7	0.020	0.65	0.64	14.0	6.6	11.0	6.3	0.140	0.47	0.62	11.4	6.4	13.4	8.0	0.395	-0.27	0.63
Lateral trunk flexion (°) - IC	-19.8	7.1	-19.4	6.5	0.830	-0.06	0.62	-18.8	6.8	-20.8	7.9	0.393	0.27	0.62	-19.1	8.3	-20.5	6.9	0.573	0.18	0.62
Lateral trunk flexion (°) - pk	-28.6	8.8	-30.5	8.1	0.478	0.22	0.62	-26.7	7.9	-33.5	9.3	0.017	0.77	0.62	-31.4	10.8	-29.0	7.2	0.413	-0.26	0.64
Lateral Trunk flexion (°) - ROM	8.7	4.9	11.1	6.5	0.204	-0.40	0.63	8.0	4.9	12.8	6.7	0.014	-0.80	0.64	12.3	5.8	8.5	5.7	0.044	0.65	0.64
Hip rotation angle (°) - IC	14.1	9.1	16.2	6.6	0.404	-0.26	0.62	12.7	8.9	13.8	7.7	0.687	-0.13	0.62	12.1	10.3	14.0	10.2	0.562	-0.18	0.62
Hip rotation angle (°) - pk	13.0	8.8	14.6	6.7	0.530	-0.20	0.62	11.5	9.0	12.4	7.2	0.732	-0.11	0.62	11.4	10.0	12.2	9.8	0.782	-0.09	0.62
Hip abduction angle (°) - IC	-23.0	4.9	-24.4	5.7	0.399	0.27	0.62	-22.0	6.0	-22.9	5.7	0.616	0.16	0.62	-22.0	5.5	-23.8	6.0	0.328	0.31	0.62
Pelvic rotation (°) - IC	31.3	8.6	28.4	9.9	0.334	0.30	0.62	32.0	9.0	27.1	7.4	0.063	0.59	0.63	28.6	8.7	31.9	8.9	0.237	-0.37	0.63
IFPA (°) - IC	17.5	8.5	13.6	12.0	0.245	0.37	0.62	19.7	8.2	12.7	8.1	0.010	0.84	0.63	13.6	8.2	18.6	12.5	0.146	-0.46	0.65
Lateral foot plant distance (m)	-0.327	0.056	-0.316	0.043	0.491	-0.23	0.62	-0.322	0.048	-0.296	0.041	0.092	-0.55	0.64	-0.294	0.045	-0.330	0.052	0.032	0.71	0.63
PFC HBF - pk (BW)	-1.49	0.37	-1.26	0.37	0.056	-0.60	0.63	-1.44	0.40	-1.50	0.41	*0.678*	0.13	0.62	-1.46	0.42	-1.39	0.37	0.583	-0.17	0.62
PFC HBF - mean (BW)	-0.59	0.07	-0.50	0.09	0.001	-1.08	0.66	-0.57	0.09	-0.55	0.10	0.580	-0.17	0.62	-0.53	0.11	-0.54	0.10	0.818	0.07	0.62
FFC VBF - pk (BW)	2.60	0.42	2.42	0.58	0.263	0.35	0.62	2.66	0.48	2.46	0.61	0.258	0.36	0.63	2.36	0.47	2.69	0.58	0.055	-0.61	0.62
FFC VBF - mean (BW)	1.62	0.12	1.56	0.25	0.343	0.30	0.62	1.68	0.20	1.53	0.21	0.031	0.70	0.64	1.51	0.19	1.68	0.23	0.016	-0.79	0.64
FFC HBF - pk (BW)	-1.41	0.29	-1.30	0.33	0.149	-0.36	0.62	-1.36	0.29	-1.37	0.36	*0.947*	0.04	0.63	-1.26	0.29	-1.43	0.32	*0.096*	0.53	0.62
FFC HBF - mean (BW)	-0.87	0.12	-0.78	0.17	0.072	-0.57	0.63	-0.88	0.16	-0.77	0.16	0.041	-0.66	0.65	-0.76	0.14	-0.88	0.16	0.010	0.84	0.64
PFC RBF - pk (BW)	2.99	0.61	2.67	0.55	0.089	0.54	0.63	2.89	0.56	3.01	0.55	0.509	-0.21	0.62	2.87	0.61	2.82	0.57	0.754	0.09	0.62
PFC RBF - mean (BW)	1.12	0.10	1.07	0.16	0.478	0.35	0.62	1.12	0.11	1.13	0.16	0.864	-0.05	0.62	1.08	0.16	1.05	0.12	0.496	0.20	0.62
FFC RBF - pk (BW)	3.04	0.42	2.82	0.64	0.081	0.40	0.63	3.13	0.52	2.87	0.66	*0.086*	0.42	0.64	2.76	0.48	3.15	0.62	0.031	-0.70	0.63
FFC RBF - mean (BW)	2.00	0.18	1.87	0.32	0.143	0.47	0.63	2.06	0.27	1.83	0.27	0.011	0.83	0.65	1.82	0.25	2.06	0.29	0.010	-0.84	0.65
FFC VPF - pk (BW)	1.87	0.21	1.76	0.28	0.043	0.46	0.63	1.97	0.27	1.69	0.17	*<0.001*	1.21	0.63	1.77	0.19	1.86	0.23	0.183	-0.42	0.67
FFC VPF - mean (BW)	1.24	0.11	1.17	0.10	0.063	0.69	0.64	1.27	0.10	1.14	0.09	*<0.001*	1.30	0.62	1.20	0.10	1.22	0.10	0.407	-0.25	0.68
FFC HPF - pk (BW)	-0.83	0.19	-0.69	0.20	0.018	-0.70	0.64	-0.88	0.21	-0.66	0.16	0.001	-1.11	0.64	-0.70	0.15	-0.82	0.19	0.036	0.68	0.67
FFC HPF - mean (BW)	-0.39	0.12	-0.30	0.08	0.007	-0.84	0.65	-0.40	0.12	-0.30	0.08	*0.002*	-0.96	0.63	-0.33	0.07	-0.37	0.09	0.144	0.46	0.65
FFC MLPF - pk (BW)	1.18	0.15	1.00	0.17	<0.001	1.05	0.66	1.22	0.16	0.95	0.12	<0.001	1.83	0.63	1.04	0.12	1.12	0.18	0.109	-0.51	0.74
FFC MLPF - mean (BW)	0.82	0.08	0.69	0.08	<0.001	1.71	0.72	0.83	0.09	0.67	0.08	<0.001	1.97	0.63	0.73	0.08	0.78	0.11	0.099	-0.52	0.75
RPF - pk (BW)	2.33	0.31	2.11	0.36	0.009	0.64	0.64	2.44	0.35	2.02	0.23	<0.001	1.38	0.63	2.14	0.24	2.29	0.32	0.091	-0.54	0.69
RPF- mean (BW)	1.56	0.15	1.41	0.13	0.001	1.06	0.66	1.59	0.15	1.37	0.12	*<0.001*	1.58	0.63	1.46	0.13	1.52	0.15	0.191	-0.42	0.71
pk HBF ratio	0.98	0.23	1.10	0.38	0.512	-0.35	0.62	1.00	0.34	0.96	0.27	0.621	0.15	0.63	0.92	0.31	1.08	0.32	*0.049*	-0.51	0.62
mean HBF ratio	1.49	0.25	1.63	0.54	0.738	-0.33	0.62	1.61	0.51	1.43	0.31	*0.253*	0.44	0.63	1.48	0.41	1.72	0.57	*0.086*	-0.48	0.63
PFC GCT (s)	0.189	0.028	0.206	0.046	0.301	-0.45	0.63	0.183	0.026	0.215	0.039	0.003	-0.96	0.63	0.212	0.037	0.193	0.036	0.116	0.50	0.65
FFC GCT (s)	0.294	0.045	0.339	0.069	0.020	-0.75	0.64	0.286	0.043	0.352	0.060	<0.001	-1.25	0.65	0.341	0.057	0.299	0.043	0.012	0.82	0.68
Approach velocity (m/s)	4.46	0.28	4.12	0.32	0.001	1.12	0.67	4.48	0.32	4.17	0.33	0.005	0.94	0.64	4.16	0.30	4.38	0.31	0.028	-0.71	0.65
FFC touch-down velocity (m/s)	3.42	0.24	3.17	0.36	0.012	0.81	0.64	3.46	0.29	3.08	0.30	<0.001	1.28	0.67	3.07	0.31	3.43	0.26	<0.001	-1.21	0.68
EXIT velocity (m/s)	3.41	0.20	3.05	0.25	<0.001	1.54	0.71	3.50	0.13	2.94	0.12	*<0.001*	4.32	0.66	3.13	0.22	3.35	0.22	0.002	-1.02	1.13
Δ PFC velocity (m/s)	-1.04	0.19	-0.96	0.28	0.256	-0.35	0.62	-1.02	0.22	-1.09	0.21	0.293	0.33	0.63	-1.08	0.20	-0.95	0.23	*0.040*	-0.60	0.62
Δ FFC velocity (m/s)	-0.83	0.22	-0.92	0.34	0.317	0.31	0.62	-0.80	0.23	-0.97	0.31	0.057	0.61	0.62	-0.85	0.28	-0.91	0.20	0.478	0.22	0.63

Small ES	0.20-0.59	Moderate ES	0.60-1.19	Large ES	1.20-1.99	Very Large	2.00-3.99	Extremely large	≥4.00

Key: pk: peak; GCT: Ground contact time; PFC: Penultimate foot contact; FFC: Final foot contact; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFM: Ankle-dorsi flexion moment; HFA: Hip flexion angle; KFA: Knee flexion angle; ADFA: Ankle dorsi-flexion angle; RPF: Resultant propulsive force; RBF: Resultant braking force; VPF: Vertical propulsive force: VBF: Vertical braking force; HPF: Horizontal propulsive force; HBF: Horizontal braking force; MLPF: Medio-lateral propulsive force; ROM: Range of motion; IC: Initial contact; BW: Body weight. Italic denotes non-parametric.

Long cut

Variable	Completion time							Exit velocity							peak KAM						
	Fast (n = 20)		Slow (n = 20)		p	g	95% CI	Fast (n = 20)		Slow (n = 20)		p	g	95% CI	Low (n = 20)		High (n = 20)		p	g	95% CI
	Mean	SD	Mean	SD				Mean	SD	Mean	SD				Mean	SD	Mean	SD			
Completion time (s)	1608	0.046	1910	0.074	*<0.001*	-4.80	122	1642	0.074	1893	0.091	*<0.001*	-2.97	0.90	1844	0.122	1695	0.119	*<0.001*	122	0.67
FFC pk HFM (Nm/kg)	-2.90	0.84	-2.51	0.79	0.138	-0.47	0.63	-2.67	0.60	-2.54	0.76	0.566	-0.18	0.62	-2.75	0.76	-2.86	0.81	0.638	0.15	0.62
FFC pk KFM (Nm/kg)	3.57	0.84	3.12	0.55	0.053	0.62	0.63	3.79	0.65	3.04	0.52	0.000	126	0.68	3.15	0.63	3.68	0.77	*0.023*	-0.74	0.64
FFC pk ADFM (Nm/kg)	-2.04	0.55	-170	0.44	0.038	-0.66	0.64	-2.05	0.55	-158	0.34	0.003	-0.99	0.66	-173	0.47	-196	0.56	0.149	0.45	0.63
FFC HFA (˚) - pk	42.7	8.4	50.7	12.4	0.022	-0.74	0.64	419	8.4	52.6	116	0.002	-104	0.66	50.8	11.2	42.9	10.3	0.026	0.72	0.64
FFC HFA (˚) - IC	40.6	7.9	43.8	8.8	0.231	-0.38	0.63	40.3	8.3	44.9	7.7	0.077	-0.57	0.63	43.7	7.4	40.6	9.0	0.235	0.38	0.63
FFC HFA (˚) - ROM	2.1	2.5	6.9	5.4	*0.003*	-1.11	0.67	16	14	7.6	5.5	*<0.001*	-147	0.70	7.0	5.3	2.3	2.8	*0.012*	109	0.66
FFC KFA (˚) - pk	59.2	5.6	62.3	8.5	0.188	-0.41	0.63	610	5.2	63.8	7.4	0.182	-0.42	0.63	63.3	7.4	610	6.7	0.305	0.32	0.62
FFC KFA (˚) - IC	22.6	5.2	22.1	4.0	0.709	0.12	0.62	23.5	4.6	22.3	3.9	0.379	0.28	0.62	218	3.7	23.6	4.9	0.209	-0.40	0.63
FFC KFA (˚) - ROM	36.6	6.7	40.2	7.2	0.110	-0.51	0.63	37.5	6.9	414	7.1	0.082	-0.55	0.63	415	7.4	37.4	5.6	0.057	0.61	0.63
FFC ADFA (˚) - pk	78.5	9.6	77.5	6.9	*0.779*	0.12	0.62	80.8	9.1	76.1	7.0	0.091	0.57	0.63	76.6	7.7	77.1	6.9	0.815	-0.07	0.62
FFC ADFA (˚) - IC	52.3	10.4	57.8	12.3	0.131	-0.48	0.63	54.6	9.2	57.1	10.1	0.422	-0.25	0.62	55.7	9.5	52.7	10.8	0.369	0.28	0.62
FFC ADFA (˚) - ROM	26.3	10.1	19.7	14.5	0.105	0.52	0.63	26.2	10.3	18.9	12.7	*0.055*	0.61	0.63	20.9	12.1	24.4	8.6	0.303	-0.32	0.62
pk KAM (Nm/kg)	127	0.36	0.89	0.37	*0.001*	101	0.66	125	0.31	0.92	0.41	0.006	0.90	0.65	0.68	0.16	156	0.21	<0.001	-4.60	118
pk KAA (˚)	-11.1	6.0	-9.9	7.4	0.589	-0.17	0.62	-11.6	6.4	-10.2	7.8	0.516	-0.20	0.62	-8.5	5.1	-15.5	5.8	<0.001	126	0.68
KAA (˚) - IC	2.8	5.1	2.8	4.7	0.987	0.00	0.62	2.1	5.2	3.5	5.6	0.422	-0.25	0.62	3.9	5.0	0.3	4.7	0.023	0.74	0.64
pk KIRM (Nm/kg)	-121	0.45	-0.66	0.35	*<0.001*	-135	0.69	-1.12	0.47	-0.64	0.27	*<0.001*	-121	0.67	-0.64	0.32	-120	0.46	*<0.001*	138	0.69
pk KRA (˚)	-3.2	8.9	-4.5	9.5	*0.652*	0.14	0.62	-2.1	9.4	-5.7	10.4	0.260	0.35	0.62	-4.6	9.5	-14	7.5	0.241	-0.37	0.63
KRA (˚) - IC	-2.4	9.9	-3.2	10.3	0.802	0.08	0.62	-13	10.4	-4.3	11.3	0.385	0.27	0.62	-3.6	9.9	10	8.7	0.134	-0.48	0.63
PFC Sag trunk inclination angle - IC (˚)	12.9	5.4	9.1	4.8	*0.009*	0.74	0.64	10.9	5.1	8.2	5.1	0.024	0.54	0.63	10.3	5.3	9.7	4.3	0.671	0.13	0.62
FFC Sag trunk inclination angle - IC (˚)	15.0	6.9	12.7	7.4	*0.301*	0.31	0.62	12.0	6.0	11.1	7.7	0.565	0.13	0.62	14.1	7.5	12.1	6.5	0.367	0.28	0.62
Lateral trunk flexion (˚) - IC	-14.2	7.7	-15.3	8.0	0.660	0.14	0.62	-14.2	6.8	-15.5	6.4	0.518	0.20	0.62	-15.9	8.8	-15.1	8.5	0.764	-0.10	0.62
Lateral trunk flexion (˚) - pk	-23.7	9.4	-27.5	11.1	0.248	0.36	0.62	-23.4	6.4	-29.2	9.7	0.032	0.69	0.64	-29.1	12.7	-24.7	9.5	0.216	-0.39	0.63
Lateral Trunk flexion (˚) - ROM	9.5	4.8	12.2	6.5	0.141	-0.47	0.63	9.3	3.7	13.7	6.7	0.015	-0.80	0.64	13.2	6.8	9.6	4.9	*0.121*	0.60	0.63
Hip rotation angle (˚) - IC	12.7	11.3	16.7	11.1	*0.132*	-0.35	0.62	12.8	11.5	17.5	8.5	0.149	-0.46	0.63	13.3	10.0	16.4	11.6	0.365	-0.28	0.62
Hip rotation angle (˚) - pk	11.3	11.0	15.1	10.4	0.255	-0.36	0.62	11.2	11.2	15.4	7.3	0.167	-0.44	0.63	12.2	9.3	14.3	11.0	0.505	-0.21	0.62
Hip abduction angle (˚) - IC	-24.9	6.7	-23.5	6.0	0.489	-0.22	0.62	-23.9	6.8	-24.8	6.7	0.652	0.14	0.62	-23.0	6.2	-25.6	7.7	0.241	0.37	0.63
Pelvic rotation (˚) - IC	37.4	7.9	31.2	10.0	0.037	0.67	0.64	35.4	7.7	30.3	9.3	0.070	0.58	0.63	32.6	8.4	35.9	10.0	0.266	-0.35	0.62
IFPA (˚) - IC	22.6	8.8	13.1	10.3	0.003	0.97	0.66	22.6	8.7	12.5	11.6	0.003	0.97	0.66	15.6	11.3	19.8	8.6	0.199	-0.41	0.63
Lateral foot plant distance (m)	-0.326	0.035	-0.303	0.058	0.132	-0.49	0.63	-0.322	0.040	-0.297	0.057	0.068	-0.50	0.63	-0.285	0.052	-0.322	0.049	0.027	0.72	0.64
PFC HBF - pk (BW)	-185	0.38	-155	0.42	0.023	-0.74	0.64	-179	0.42	-164	0.38	*0.355*	-0.36	0.62	-159	0.32	-161	0.45	*1000*	0.06	0.62
PFC HBF - mean (BW)	-0.68	0.09	-0.52	0.08	<0.001	-181	0.74	-0.66	0.11	-0.55	0.09	0.002	-101	0.66	-0.57	0.11	-0.61	0.13	0.269	0.35	0.62
FFC VBF - pk (BW)	2.74	0.48	2.51	0.57	*0.072*	0.43	0.63	2.62	0.47	2.39	0.47	0.096	0.47	0.63	2.39	0.55	2.63	0.39	*0.049*	-0.49	0.63
FFC VBF - mean (BW)	166	0.17	151	0.17	0.007	0.89	0.65	164	0.17	147	0.15	0.001	107	0.66	148	0.16	167	0.18	*0.002*	-1.10	0.67
FFC HBF - pk (BW)	-156	0.33	-142	0.35	*0.127*	-0.43	0.63	-149	0.32	-138	0.32	*0.149*	-0.36	0.62	-137	0.36	-147	0.27	*0.091*	0.31	0.62
FFC HBF - mean (BW)	-0.96	0.15	-0.77	0.13	<0.001	-125	0.68	-0.91	0.13	-0.77	0.13	0.001	-108	0.66	-0.78	0.16	-0.93	0.16	0.005	0.91	0.65
PFC RBF - pk (BW)	3.30	0.56	2.88	0.58	0.026	0.72	0.64	3.28	0.61	2.98	0.55	0.103	0.52	0.63	2.91	0.55	3.11	0.69	0.332	-0.30	0.62
PFC RBF - mean (BW)	123	0.15	110	0.17	0.014	0.80	0.64	124	0.19	113	0.15	0.043	0.65	0.64	114	0.15	116	0.20	0.618	-0.16	0.62
FFC RBF - pk (BW)	3.27	0.54	2.99	0.64	0.148	0.46	0.63	3.11	0.52	2.84	0.54	*0.115*	0.50	0.63	2.84	0.64	3.14	0.52	*0.086*	-0.49	0.63
FFC RBF - mean (BW)	2.08	0.23	182	0.22	0.001	112	0.67	2.04	0.22	177	0.20	0.000	124	0.68	179	0.23	2.06	0.25	0.001	-1.11	0.67
FFC VPF - pk (BW)	195	0.26	171	0.28	0.007	0.88	0.65	194	0.24	163	0.21	*<0.001*	136	0.69	177	0.28	190	0.31	0.198	-0.41	0.63
FFC VPF - mean (BW)	125	0.09	115	0.12	*0.001*	0.99	0.66	124	0.08	112	0.09	*<0.001*	141	0.69	119	0.14	122	0.12	0.484	-0.22	0.62
FFC HPF - pk (BW)	-0.96	0.20	-0.69	0.21	<0.001	-130	0.68	-0.91	0.18	-0.65	0.19	<0.001	-137	0.69	-0.74	0.24	-0.89	0.23	0.042	0.65	0.64
FFC HPF - mean (BW)	-0.44	0.11	-0.31	0.10	<0.001	-120	0.67	-0.40	0.08	-0.30	0.10	*0.002*	-102	0.66	-0.35	0.12	-0.40	0.13	0.229	0.38	0.63
FFC MLPF - pk (BW)	123	0.18	0.98	0.15	<0.001	150	0.70	122	0.17	0.93	0.13	<0.001	182	0.74	102	0.15	114	0.26	0.070	-0.58	0.63
FFC MLPF - mean (BW)	0.84	0.08	0.68	0.09	<0.001	182	0.74	0.83	0.08	0.66	0.08	<0.001	2.19	0.78	0.71	0.09	0.77	0.14	0.150	-0.47	0.63
RPF - pk (BW)	2.47	0.37	2.06	0.36	0.001	110	0.67	2.44	0.33	197	0.28	<0.001	150	0.70	2.15	0.37	2.36	0.45	0.113	-0.50	0.63
RPF- mean (BW)	159	0.14	138	0.16	<0.001	137	0.69	156	0.11	135	0.13	*<0.001*	175	0.73	144	0.18	151	0.20	0.267	-0.35	0.62
pk HBF ratio	0.87	0.21	0.95	0.24	0.256	-0.36	0.62	0.87	0.22	0.87	0.24	0.913	-0.03	0.62	0.89	0.23	0.96	0.25	*0.478*	-0.30	0.62
mean HBF ratio	142	0.28	152	0.32	0.323	-0.31	0.62	141	0.25	141	0.30	*0.718*	-0.02	0.62	138	0.22	157	0.40	*0.127*	-0.56	0.63
PFC GCT (s)	0.177	0.034	0.225	0.044	*<0.001*	-1.18	0.67	0.182	0.032	0.225	0.041	0.001	-1.15	0.67	0.217	0.039	0.184	0.034	*0.003*	0.89	0.65
FFC GCT (s)	0.265	0.036	0.336	0.070	*<0.001*	-126	0.68	0.272	0.034	0.348	0.065	<0.001	-145	0.70	0.333	0.068	0.289	0.045	*0.021*	0.74	0.64
Approach velocity (m/s)	4.94	0.34	4.28	0.28	*<0.001*	2.06	0.77	4.89	0.38	4.33	0.31	<0.001	159	0.71	4.38	0.33	4.71	0.37	*0.004*	-0.94	0.65
FFC touch-down velocity (m/s)	3.81	0.30	3.15	0.19	<0.001	2.57	0.84	3.79	0.30	3.14	0.19	<0.001	2.54	0.83	3.22	0.29	3.62	0.30	*<0.001*	-135	0.69
EXIT velocity (m/s)	3.56	0.24	3.03	0.18	<0.001	2.42	0.82	3.64	0.17	2.96	0.12	*<0.001*	4.52	117	3.12	0.23	3.43	0.34	0.002	-105	0.66
Δ PFC velocity (m/s)	-113	0.17	-113	0.21	0.980	0.01	0.62	-111	0.19	-119	0.19	0.149	0.46	0.63	-116	0.16	-109	0.19	0.215	-0.39	0.63
Δ FFC velocity (m/s)	-0.86	0.18	-0.92	0.28	*0.640*	0.23	0.62	-0.84	0.16	-0.98	0.25	0.042	0.65	0.64	-0.89	0.22	-0.89	0.18	0.932	-0.03	0.62
Approach time (s)	1877	0.087	2.090	0.170	*<0.001*	-154	0.71	1903	0.113	2.090	0.172	<0.001	-126	0.68	1991	0.152	1956	0.160	0.489	0.22	0.62
	Small ES		0.20-0.59			Moderate ES		0.60-1.19		Large ES		1.20-1.99		Very Large		2.00-3.99		Extremely large			≥4.00

Key: pk: peak; GCT: Ground contact time; PFC: Penultimate foot contact; FFC: Final foot contact; KAM: Knee abduction moment; KIRM: Knee internal rotation moment; KFM: Knee flexion moment; HFM: Hip flexion moment; ADFM: Ankle-dorsi flexion moment; HFA: Hip flexion angle; KFA: Knee flexion angle; ADFA: Ankle dorsi-flexion angle; RPF: Resultant propulsive force; RBF: Resultant braking force; VPF: Vertical propulsive force: VBF: Vertical braking force; HPF: Horizontal propulsive force; HBF: Horizontal braking force; MLPF: Medio-lateral propulsive force; ROM: Range of motion; IC: Initial contact; BW: Bodyweight. Italic denotes non-parametric.

Appendix 5.1 Cutting movement assessment score: CMAS operational definitions and biomechanical rationale (Chapter 6)

Suggested viewing camera	CMAS variable	Observation and score	Operational Definition	Biomechanical rationale
Penultimate contact				
Side / 45°	Clear PFC braking strategy (at initial contact) Backward inclination of the trunk Large COM to COP position - anterior placement of the foot Effective deceleration – heel contact PFC	Y/N Y=0/ N=1	If the subject does not demonstrate a clear PFC braking strategy that emphasises large anterior placement of the foot relative to the centre of mass and does not demonstrate backwards trunk inclination (relative to a vertical straight line), then a score (+1) is awarded Practitioners may consider referring to a vertical straight line for evaluating trunk inclination.	COD is multistep action with evidence to suggest that the PFC is involved in deceleration prior to directional change, and is a 'preparatory step' (11, 273). A 'large anterior placement of the foot relative to COM and backward inclination of the trunk relative to planted foot is considered to increase horizontal braking forces during PFC, based on research demonstrating a relationship between average horizontal GRF during PFC and peak KAMs during FFC (273). Reducing the majority of momentum during PFC, will reduce the braking requirements of the FFC, which may result in lower knee joint loads and protect against injury (272-274).
Final Contact				
Front / 45°	Wide lateral leg plant (approx. > 35 cm – dependent on subject anthropometrics) (at initial contact)	Y/N Y=2/N=0	If the subject demonstrates a wide lateral foot leg plant, then a score (+2) is awarded. A wide foot plant is considered as a distance > 0.35 m) between the hip and plant foot contact IC; however, this is dependent on subject anthropometrics. Practitioners assessing athletes of small stature and leg length may change this accordingly (i.e., > 0.25 m for youth athletes)	A 'wide lateral leg plant' is a major determinant of peak KAM (127, 228, 272, 301). A wide foot plant creates a GRF vector acting laterally relative to the knee, whereby greater foot plant distances creating a greater moment arm (relative to knee joint centre) and thus, KAM. Abducted hip positions are also commonly observed characteristics displayed during visual inspection of non-contact ACL injuries during COD actions (63, 204, 270, 291).
Front / 45°	Hip in an initial internally rotated position (at initial contact)	Y/N Y=1/N=0	If the subject's femur is in an internally rotated position at initial contact, then a score (+1) is awarded.	Internal hip rotation can lead to a more medially positioned knee relative to the GRF vector, thus increase moment arm distance and subsequent KAM (228, 358, 516, 519).
Front / 45°	Initial knee 'valgus' position (at initial contact)	Y/N Y=1/N=0	If the subject's knee is in a valgus (medial) position at initial contact, then a score (+1) is awarded.	'Initial knee valgus position' has been shown to be associated with peak KAM (272, 274, 301, 358, 516). An increased knee abduction angle at initial contact has an effect of placing the knee more medial to the resultant GRF vector and thus, increases the lever arm of the resultant GRF vector relative to the knee joint, leading to an increased KAM. Prospective research showed greater valgus angles were associated with increased risk of non-contact ACL injury (242). Increase in knee valgus angle of 2° can lead to a 40 Nm change in valgus moment (359).
All 3	Foot not in neutral foot position (at initial contact) Inwardly rotated foot position or externally rotated foot position (relative to approach direction of travel)	Y/N Y=1/N=0	If the subject's foot is not in a neutral position (i.e., approx. 0°) and is inwardly or externally rotated at initial contact (relative to approach direction of travel), then a score (+1) is awarded	Initial foot progression angle is associated with KAM, with a neutral foot position considered the safest strategy (127, 274, 519). Internally rotated foot positions during weight acceptance can lead to a more medially positioned knee relative to the GRF vector, thus increase moment arm distance and subsequent KAM (274, 519). A neutral foot position would most likely result in forces being absorbed in the sagittal plane utilising the large knee and hip extensor musculature, which is potentially a safer strategy (274). Excessive foot external rotation increases susceptibility to eversion and pronation which could lead to knee valgus and tibial internal rotation (180, 334, 419), thus ACL loading. External rotation of the foot has

				also been stated as characteristics during visual inspection of non-contact ACL injuries during change of direction (270).
Front / 45°	Frontal plane trunk position relative to intended direction; Lateral or trunk rotated towards stance limb (L/TR), Upright (U) or Medial (M) (at initial contact and over WA) (use shoulder positioning as guide)	L/TR/U/M L=2/ TR = 2/ U = 1, /M=0	If the subject's trunk is laterally flexed over the stance (push-off) limb or rotated towards the stance limb at initial contact and over WA, then a score (+2) is awarded. If the subject's trunk is upright (vertical relative to straight line), then a score (+1) is awarded. If the subject's trunk is medial (leaning towards the intended direction of travel), then no score is awarded. Practitioners may consider referring to a vertical straight line and use shoulder position as an indicator.	The trunk contains approx. half of the body's mass, and during cutting the entire body's mass must be balanced and supported on one leg, thus trunk control and positioning is a critical factor influencing knee joint loads (245, 248, 366). Lateral trunk flexion (127, 267, 272) or trunk rotation (127, 189) towards stance limb are major determinants of peak KAM. A laterally flexed trunk or rotated trunk towards the planted leg side shifts the athlete's weight laterally creating a laterally directed force vector, increasing the moment arm relative to the knee joint and thus, KAMs. Prospective research has shown deficits in trunk control and proprioception are associated with increased risk of non-contact ACL injury (635, 636). Lateral trunk flexion over plant leg also a commonly observed visual characteristic of non-contact ACL injuries during plant-and cut manoeuvres and landing (248, 291, 541).
Side / 45°	Trunk upright or leaning back throughout contact over whole contact (not adequate trunk flexion displacement) (at initial contact and over WA)	Y/N Y=1/N=0	If the subject's trunk is upright or leaning back throughout weight (i.e., appears limited hip flexion) acceptance and push-off during the FFC and does not go through an adequate range of trunk-flexion displacement, then a score (+1) is awarded.	Trunk inclination (leaning back or upright) with minimal trunk flexion displacement during weight acceptance may increase the overall knee joint load due an increased lever arm of the trunk relative to the knee and increasing the COM distance from the base of support (509). Some trunk flexion allows generation of hip moments to help absorb the GRF during weight acceptance and thus, may lower KAMs (274). Increasing hip flexion and promoting a hip dominant strategy are involved GRF attenuation (511, 517, 631), energy dissipation (459, 624), reducing loading rates (517) and reducing knee joint loads (406, 459, 464, 517) during high impact tasks. Increasing hip flexion increases the moment arm distance at the hip which creates a greater hip flexor moment (utilising hip extensor musculature). This can have the effect of unloading the knee by more evenly distributing loading proximally up the lower-limb chain (274, 459, 464, 517), thus reducing the demands for the knee.
All 3	Limited knee flexion during final contact (stiff) ≤ 30˚	Y/N Y=1/N=0	If the subject's knee goes through limited knee flexion (approximately ≤ 30˚) over weight acceptance appears 'stiff', then a score (+1) is awarded.	Stiffer weight acceptance strategies can increase impact GRFs (115, 130, 640), and greater GRFs are associated with increased KAMs (516, 519). Less knee flexion is also associated with greater KAMs (301, 593). Furthermore, extended knee positions with high anterior tibial loading and shear force can also increase ACL strain (42, 43, 343, 615). Extended knee positions are also commonly observed characteristics of non-contact ACL injury during directional changes and landing (52, 63, 94, 204, 270, 291, 298, 310, 376, 426, 583).
Front / 45°	Excessive knee 'valgus' motion during WA	Y/N Y=1/N=0	If the subject's knee demonstrates visible valgus motion during weight acceptance, then a score (+1) is awarded.	Knee valgus motion during FFC is considered because it is a key indicator of ACL injury risk (242), and can contribute to large front plane knee joint loading. Dynamic knee valgus positions are also commonly observed characteristics of non-contact ACL injury during directional changes and landing (63, 94, 204, 270, 291, 298, 310, 426, 541, 583).
Key: PFC: Penultimate foot contact; FFC: Final foot contact; KAM: peak Knee abduction moment: ACL: Anterior cruciate ligament; IC: Initial contact; WA: Weight acceptance; GRF: Ground reaction force. Y: Yes; N: No; L: Lateral; TR: Trunk rotation; U: Upright; M: Medial.				

Appendix 5.2 CMAS high- and low-risk images (Chapter 6)

Table 1. Cutting movement assessment score tool

Camera	Variable	Observation	Score
<u>Penultimate contact</u>			
Side / 45°	**Clear PFC braking strategy** (at initial contact) • Backward inclination of the trunk • Large COM to COP position – anterior placement of the foot • Effective deceleration – heel contact PFC	Y/N	Y=0/ N=1

Yes

No

If the subject does not demonstrate a clear PFC braking strategy that emphasises large anterior placement of the foot relative to the centre of mass and does not demonstrate backwards trunk inclination (relative to a vertical straight line), then a score (+1) is awarded.

Practitioners may consider referring to a vertical straight line for evaluating trunk inclination.

Table 1. Cutting movement assessment score tool

Camera	Variable	Observation	Score
<u>Final Contact</u>			
Front / 45°	**Hip in an initial internally rotated position** (at initial contact)	Y/N	Y=1/N=0

Yes

No

If the subject's femur **is in an internally rotated position at initial contact, then a score (+1) is awarded.**

Table 1. Cutting movement assessment score tool

Camera	Variable	Observation	Score
<u>Final Contact</u>			
Front / 45°	**Wide lateral leg plant** (approx. > 0.35 m – dependent on subject anthropometrics) (at initial contact)	Y/M/ N	Y=2/ M= 1/N=0

Yes

Mod

No

If the subject demonstrates a wide lateral foot leg plant, then a score (+2) is awarded. A wide foot plant is considered as the distance > 35 cm between the hip and plant foot contact IC; however, this is dependent on subject anthropometrics. Practitioners should consider the angle of the tibia – approx. 45° is wide. Near vertical is narrow. Practitioners assessing athletes of small stature and leg length raters may change this accordingly (i.e. 25 cm for children).

Table 1. Cutting movement assessment score tool

Camera	Variable	Observation	Score
<u>Final Contact</u>			
Front / 45°	**Initial knee 'valgus' position** (at initial contact)	Y/N	Y=1/N=0

Yes

No

If the subject's knee is in a valgus (medial) position at initial contact, then a score (+1) is awarded.

Table 1. Cutting movement assessment score tool

Camera	Variable	Observation	Score
Final Contact			
All 3	**Foot not in neutral foot position** (@ initial contact) Inwardly rotated foot position or externally rotated foot position (relative to original direction of travel)	Y/N	Y=1/N=0

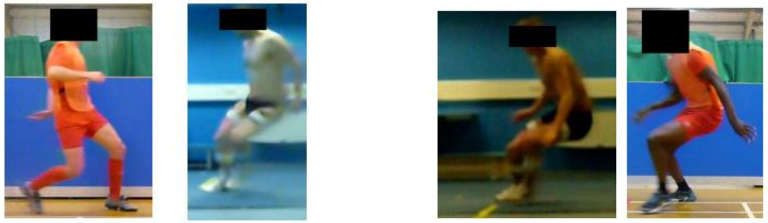
Yes **No**

If the subject's foot is not in a neutral position (i.e. approx. 0˚) and is inwardly or externally rotated at initial contact (relative to approach direction of travel), then a score (+1) is awarded

Table 1. Cutting movement assessment score tool

Camera	Variable	Observation	Score
Final Contact			
Side / 45˚	**Trunk upright or leaning back throughout contact** (not adequate trunk flexion displacement) - (at initial contact and over WA)	Y/N	Y=1/N=0

Yes **No**



If the subject's trunk is upright or leaning back throughout weight acceptance and push-off during the FFC and does not go through an adequate range of trunk-flexion displacement, then a score (+1) is awarded.

Table 1. Cutting movement assessment score tool

Camera	Variable	Observation	Score
Final Contact			
Front / 45˚	**Frontal plane trunk position relative to intended direction; Lateral or trunk rotated towards stance limb, Upright or Medial.** (at initial contact and over WA)	L/TR/U/M	L/TR=2/ U = 1, /M=0

L **TR** **U** **M**

If the subject's trunk is laterally flexed or over the stance (push-off) limb or rotated towards the stance limb at initial contact and over WA, then a score (+2) is awarded.

If the subject's trunk is upright (vertical relative to straight line), then a score (+1) is awarded. If the subject's trunk is medial (facing towards the intended direction of travel), then no score is awarded. Practitioners may consider referring to a vertical straight line and use shoulder position as an indicator.

Table 1. Cutting movement assessment score tool

Camera	Variable	Observation	Score
Final Contact			
Side / 45˚	**Limited Knee Flexion during final contact (stiff) ≤ 30˚** (over WA)	Y/N	Y=1/N=0

Yes **No**

If the subject's knee goes through limited knee flexion (approximately ≤ 30˚) and weight acceptance appears 'stiff', then a score (+1) is awarded.

Table 1. Cutting movement assessment score tool

Camera	Variable	Observation	Score
Final Contact			
Front / 45°	**Excessive Knee 'valgus' motion during contact** (over WA)	Y/N	Y=1/N=0

Yes **No**




If the subject's knee visibly demonstrates valgus motion during weight acceptance, then a score (+1) is awarded.

Appendix 6.1 Six-week COD speed and technique modification intervention programme (Chapter 7)

		Session 1				Session 2			
Week	**COD Emphasis**	**Drills**	**Intensity (perceived speed)**	**Total Distance (m)**	**Number of Decelerations and CODs**	**Drills**	**Intensity (perceived speed)**	**Total Distance (m)**	**Number of Decelerations and CODs**
Week 1	• Drills specific to deceleration phase, before adding turn and reacceleration • Submaximal/ pre-planned emphasising key aspects of technique • Progressive increase in COD angle and approach velocity	1. 10 × 5-m acceleration to deceleration (2 reps a go) 2. 12 × 5-m acceleration to side-steps (20-60˚) – 5-m exit to deceleration – 2 reps at 20°, 40°, 60° 3. 8 × 5-m acceleration to 135˚ turns – 5-m exit to deceleration	50-75%	250	25 and 20	1. 10 × 5 m acceleration to deceleration (2 reps a go) 2. 12 × 5 m acceleration to side-steps (20-60˚) – 5-m exit to deceleration – 2 reps at 20°, 40°, 60° 3. 8 × 5-m acceleration to 135˚ turns – 5-m exit to deceleration	50-75%	250	25 and 20
Week 2		1. 8 × 5-m acceleration to deceleration 2. 14 × 5-m acceleration to side steps (45-90˚) – 5-m exit to deceleration 3. 8 × 5-m acceleration to 135-180˚ turns – 5-m exit to deceleration	75%+	230	30 and 22	1. 4 × 5-m acceleration to deceleration 2. 4 × 10-m acceleration to deceleration 3. 3 × 80° zig-zag runs – (2 cuts a rep)- 5-m between cuts to 5-m deceleration (60% effort) 4. 4 × 80° zig-zag runs – (2 cuts a rep) – 5-m between cuts to 5-m deceleration 5. 2 × 80° zig-zag runs with 180° turn– 4 cuts per trial – 5-m between cuts 6. 3 × 5-m approach to 180°, 5-m approach to 180°, to 10-m deceleration	100%	285	17 and 30
Week 3	• Pre-planned drill performed maximally • Introduction of unanticipated generic stimuli (auditory or visual) • Unanticipated performed submaximally	1. 2 × 5 reps per limb: Jump-landings – lateral reactive jumps 2. 4 × 10-m acceleration to deceleration 3. 5 × 45° zig-zag runs (2 cuts a rep) -5-m between cuts 4. 10 × reactive y-agility (45° cut) – 5-m entry and exit*	100%	215	14 and 20	1. 2 × 5-m acceleration to deceleration- 2. 4 × 2.5-10 m unanticipated decelerations – auditory stimuli 3. 2 × 15-m zig-zag runs (2 90˚ cuts) – 5-m between cuts to 5-m deceleration 4. 2 × 15-m zig-zag runs relay (2 90˚ cuts) – 5-m between cuts to 5-m deceleration 5. 8 × reactive y-agility (60° cut) – 7.5-m entry and exit* 6. 3 × 5-m approach to 180°, 5-m approach to 180°, to 10-m deceleration	100%	280-310	11 and 23
Week 4		1. 2 × 5 reps per limb: Jump-landings – lateral reactive jumps 2. 4 × 20-m acceleration to deceleration 3. 4 × 45° zig-zag runs (2 cuts a rep) -6-m between cuts 4. 4 × 15-m tunnel drill – side-stepping past 2 opponents every 7.5-m 5. 4 × unanticipated 45° cuts (auditory stimuli) – 7.5-m entry and exit 6. 6 × reactive y-agility (45° cut) – 5-m entry and exit*	1. 100% 2. 75%+ 3. 100% 4. 75%+ 5. 75%+ 6. 100%	282	4 and 32	1. 4 × 5-15 m unanticipated decelerations – partner stimuli* 2. 2 × 15-m zig-zag runs (2 90˚ cuts) – 5-m between cuts to 5-m deceleration 3. 2 × 30-m zig-zag runs relay (4 90˚ cuts) – 5-m between cuts to 5-m deceleration 4. 4 × unanticipated 45° cuts (auditory stimuli) – 5-m entry and exit 5. 6 × reactive y-agility (60° cut) 2 cuts – 5-m entry and exit* 6. 4 × 15-m tunnel drill – side-stepping past 2 opponents every 7.5-m	100%	330-370	16 and 38

Week 5	• Unanticipated drills performed maximally • Introduction of sport specific stimuli – opponent or ball	1. 2 × 15-m curvilinear sprint (s shape) 2. 12 × 5-m acceleration to side-step (60˚) – 5-m exit to deceleration 3. 4 × 15-m tunnel drill – side-stepping past 2 opponents every 7.5-m 4. 4 × unanticipated 45° cuts (auditory stimuli) – 7.5-m entry and exit 5. 6 × reactive y-agility (60° cut) 2 cuts – 7.5m entry and exit*	100%	285	10 and 32	1. 4 x 10-15-m unanticipated decelerations – partner stimuli* 2. 2 x 180° races – two turns at any point across 15-m distance – partner lead* 3. 8 x X-agility: 1 pivot and 1 cut per rep – up to 15m* 4. 3 x 35-m zig-zag runs races (5 x 90˚ cuts) – 5-m between cuts to 5-m deceleration – 1st cut unanticipated	100%	~405-425	15 and 35
Week 6		1. 12 × 5-m acceleration to side-step (60˚) – 5-m exit to deceleration 2. 3 × Zig-zag races – 25-m length 4 x 45° cuts per race – 1st cut unanticipated 3. 4 x unanticipated 45° cuts (auditory stimuli) – 7.5-m entry and exit 4. 10 × reactive y-agility (60° cut) cuts – 7.5m entry and exit*	100%	345	48 (COD)	1. 4 × 10-15-m unanticipated decelerations – partner stimuli* 2. 2 × 180° races – two turns at any point across 15-m distance – partner lead* 3. 8 × X-agility: 1 pivot and 1 cut per rep – up to 15m* 4. 3 × 35-m zig-zag runs races (5 x 90˚ cuts) – 5-m between cuts to 5-m deceleration – 1st cut unanticipated	100%	~405-425	15 and 35

Additional information:

1. 30-60 seconds' rest provided between 100% effort reps. 2 minutes' rest provide between exercises
2. All CODs and decelerations to be performed with the aim of modified braking and COD strategy
3. Feedback to be provided to each player after each rep regarding braking strategy/ COD technique

Key: * = Alternate between leading and reacting / attacking and defending; COD = Change of direction; PFC = Penultimate foot contact; Dec: Deceleration

Appendix 7.1. COD speed and technique modification 6-week training intervention (Chapter 8)

Week	COD Emphasis	Drills	Intensity (perceived speed)	Total Distance (m)	Number of Decelerations and CODs
Week 1	• Drills specific to deceleration phase, before adding turn and reacceleration • Submaximal/ pre-planned emphasising key aspects of technique • Progressive increase in COD angle and approach velocity	1. 10 × 5-m acceleration to deceleration (2 reps a go) 2. 12 × 5-m acceleration to side-steps (20-60˚) – 5-m exit to deceleration – 2 reps at 20°, 40°, 60˚ 3. 8 × 5-m acceleration to 135° turns – 5-m exit to deceleration 4. 2 × Max effort 135° pivot each leg, max effort 60° cut each leg	50-75%	290	29 and 24
Week 2		1. 4 × 5-m acceleration to deceleration 2. 4 × 10-m acceleration to deceleration 3. 12 × 5-m acceleration to side-steps (45°, 60°, and 80˚) – 5-m exit to deceleration 4. 8 × 5-m acceleration to 135° turns – 5-m exit to deceleration 5. 2 × Combination: 5-m 60° cut, to 180° turn, to 60° cut, to Dec	75%+	300	30 and 26
Week 3	• Pre-planned drill performed maximally • Introduction of unanticipated generic stimuli (auditory or visual) • Unanticipated performed submaximally	1. 4 × 5- and 10-m acceleration to deceleration 2. 4 × 2.5-10-m acceleration to deceleration- react to coach shout 3. 16 × 5-m acceleration to side-steps (45° and 90˚) – 5-m exit to deceleration 4. 4 × 5-m combined 180˚ - 2 x 180° per rep 5. 3 × 20-m tunnel drill – side-stepping past 4 opponents every 5-m	75-100%	320-350	28 and 36
Week 4		1. 4 × 5-15 m unanticipated decelerations – partner stimuli* 2. 3 × zig-zag runs (two 45˚ cuts) – 5-m exit to deceleration (15-m total) – one partner chase 3. 3 × zig-zag runs (two 90˚ cuts) – 5-m exit to deceleration (15 m total) – one partner chase 4. 6 × unanticipated 70° cuts (coach body position) – 5-m entry and exit 5. 3 × pro-agility (5-m approach to 180°, 5-m reaccelerate to 180°, to 10-m deceleration)	100%	260-300	19 and 24
Week 5	• Unanticipated drills performed maximally • Introduction of sport specific stimuli – opponent	1. 4 × 5-15 m unanticipated decelerations – partner stimuli* 2. 3 × zig-zag runs (two 45˚ cuts) – 5-m exit to deceleration (15-m total) – one partner chase 3. 3 × zig-zag runs (two 90˚ cuts) – 5-m exit to deceleration (15-m total) – one partner chase 4. 4 × unanticipated 70° cuts (coach body position) – 5-m entry and exit 5. 4 × unanticipated 70° cuts (follow opponent) – 5-m entry and exit* 6. 2 × 180° races – two turns every 5-m to 5-m deceleration	100%	260-300	26 and 28
Week 6		1. 4 x 5-15-m unanticipated decelerations – partner stimuli- two decelerations per rep* 2. 8 x 5-m acceleration to side steps (45˚) – 5-m exit to deceleration 3. 4 × Modified L runs – anticipated - (5-m acceleration to 90˚ cut, 5-m acceleration to 180˚ turn – 5-m acceleration to 90˚ cut – to 5-m deceleration 4. 6 × X-drill – two cuts/turns in a square – partner follow* 5. 2 × 180° races – two turns at any point across 10-m distance – partner lead*	100%	270-330	28 and 36

Additional information:

4. 30-60 seconds' rest provided between 100% effort reps. 2 minutes' rest provide between exercises
5. All CODs and decelerations to be performed with the aim of modified braking and COD strategy
6. Feedback to be provided to each player after each rep regarding braking strategy/ COD technique

Key: * = Alternate between leading and reacting / attacking and defending; COD = Change of direction; PFC = Penultimate foot contact; Dec: Deceleration

Appendix 7.2. Jump-landing warm-up drills performed during COD speed and technique modification training intervention (Chapter 8)

Week	Warm-up jump-landing plyometric drills
Week 1	Jump-landing: 10-m of: 1. Bilateral broad jump 2. Bilateral broad jump – zig-zag 3. Bilateral broad jump - lateral 4. Single-leg hop and hold – forwards 5. Single-leg hop and hold – zig-zag 6. Single-leg cutting push off action and hold
Week 2	Jump-landing: 10-m of: 1. Bilateral broad jump 2. Bilateral broad jump – zig-zag 3. Bilateral broad jump - lateral 4. Single-leg hop and hold – forwards 5. Single-leg hop and hold – zig-zag 6. Single-leg cutting push off action and hold
Week 3	Jump-landing: 12-m of: 1. Bilateral broad jump (reactive in pairs) 2. Bilateral broad jump – zig-zag 3. Bilateral broad jump - lateral 4. Single-leg hop and hold – forwards 5. Single-leg hop and hold – zig-zag 6. Single-leg cutting push off action and hold
Week 4	Jump-landing: 12-m of: 1. Bilateral broad jump (reactive in pairs) 2. Forward tuck-jumps 3. Zig-zag tuck-jumps 4. Single-leg hop and hold – forwards 5. Single-leg hop and hold – zig-zag 6. Single-leg cutting push off action continuous
Week 5	Jump-landing: 12-m of: 1. Bilateral broad jump (reactive in pairs) 2. Forward tuck-jumps 3. Zig-zag tuck-jumps 4. Single-leg hop and hold – forwards – double hop 5. Single-leg hop and hold – zig-zag 6. Single-leg cutting push off action continuous 7. Crossover hop and hold
Week 6	Jump-landing: 12-m of: 1. Bilateral broad jump (reactive in pairs) 2. Forward tuck-jumps 3. Zig-zag tuck-jumps 4. Single-leg hop and hold – forwards – double hop 5. Single-leg hop and hold – zig-zag 6. Single-leg cutting push off action continuous 7. Crossover hop and hold

Appendix 8.1 List of publications

(Those in bold below are publications resulting from this thesis)

1. **Dos'Santos, T., McBurnie, A., Comfort, P., & Jones, P.A. (2019) The Effects of Six-Weeks Change of Direction Speed and Technique Modification Training on Cutting Performance and Movement Quality in Male Youth Soccer Players. *Sports*, *7*, 205.**
2. Thomas., C., Dos'Santos, T., Cuthbert, M., Fields, C., & Jones, P. A. (2019). The Effect of Limb Preference on Braking Strategy and Knee Joint Mechanics During Pivoting in Female Soccer Players. *Science and Medicine in Football*. Epub ahead of print.
3. **McBurnie, A. Dos'Santos, T., & Jones, P A. (2019). The performance-injury conflict during cutting. Journal of Strength and Conditioning Research. Epub ahead of print.**
4. **Dos' Santos, T., Comfort, P., & Jones, P. A. (2018). Average of trial peaks versus peak of average profile: impact on change of direction biomechanics. *Sports biomechanics*, 1-10, Epub ahead of print.**
5. **Jones, P.A., Donelon, T., & Dos' Santos, T. (2017). A preliminary investigation into a qualitative assessment tool to identify athletes with high knee abduction moments during cutting: Cutting Movement Assessment Score (CMAS). *Professional Strength and Conditioning*, 47, 37-42.**
6. Jones, P., Thomas, C., Dos'Santos, T., McMahon, J., & Graham-Smith, P. (2017). The role of eccentric strength in 180 turns in female soccer players. *Sports*, *5*(2), 42.
7. **Dos'Santos, T., Thomas, C., Comfort, P., & Jones, P.A. (2019). The effect of training interventions on change biomechanics associated with increased anterior cruciate ligament loading: a scoping review. *Sports Med*. Epub ahead of print.**
8. **Dos'Santos, T., McBurnie, A., Thomas, C., Comfort, P., & Jones, P. (2019). Biomechanical Comparison of Cutting Techniques: A Review and Practical Applications. *Strength & Conditioning Journal, 41*(4), 40-54.**
9. **Dos'Santos, T., Bishop, C., Thomas, C., Comfort, P., & Jones, P. A. (2019). The effect of limb dominance on change of direction biomechanics: a systematic review of its importance for injury risk. *Physical therapy in sport, 37,* 179-189.**
10. **Dos' Santos, T., Thomas, C., Comfort, P., & Jones, P. A. (2019). Role of the Penultimate Foot Contact During Change of Direction: Implications on Performance and Risk of Injury. *Strength & Conditioning Journal*, *41*(1), 87-104.**
11. **Dos'Santos, T., Thomas, C., Comfort, P., & Jones, P. A. (2018). The effect of angle and velocity on change of direction biomechanics: An angle-velocity trade-off. *Sports Medicine*, *48*(10), 2235-2253.**
12. Thomas, C., Dos'Santos, T., Comfort, P., & Jones, P. (2018). Relationships between unilateral muscle strength qualities and change of direction in adolescent team-sport athletes. *Sports*, *6*(3), 83.
13. **Dos'Santos, T., McBurnie, A., Donelon, T., Thomas, C., Comfort, P., & Jones, P. A. (2019). A qualitative screening tool to identify athletes with 'high-risk' movement mechanics during cutting: The cutting movement assessment score (CMAS). *Physical therapy in sport*, *38*, 152-161.**
14. Davidson, T.., Jarvis, P., Dos'Santos, T., Turner, T., & Bishop, C. (2019). Modifying the pro-agility test: is the change of direction deficit affected by a rolling start? *Professional strength and conditioning, 57*, 21-29.
15. Cuthbert, M., Thomas, C., Dos' Santos, T., & Jones, P. A. (2019). Application of Change of Direction Deficit to Evaluate Cutting Ability. *The Journal of Strength & Conditioning Research*, *33*(8), 2138-2144.
16. Dos'Santos, T., Thomas, C., Comfort, P., & Jones, P. A. (2018). Comparison of change of direction speed performance and asymmetries between team-sport athletes: Application of change of direction deficit. *Sports*, *6*(4), 174.
17. Dos' Santos, T., Thomas, C., Jones, P. A., & Comfort, P. (2018). Asymmetries in isometric force-time characteristics are not detrimental to change of direction speed. *The Journal of Strength & Conditioning Research*, *32*(2), 520-527.

18. Dos'Santos, T., Thomas, C., Jones, P. A., & Comfort, P. (2018). Assessing asymmetries in change of direction speed performance; application of change of direction deficit. *J. Strength Cond. Res*. Epub ahead of print.
19. Dos'Santos, T., Thomas, C., Jones, P. A., & Comfort, P. (2017). Asymmetries in single and triple hop are not detrimental to change of direction speed. *Journal of Trainology*, *6*(2), 35-41.

Poster and oral presentations

1. Dos'Santos, T., Comfort, P., & Jones, P A. (2018). Effect of low pass filtering cut-off frequencies on knee abduction moments during cutting. BASES BIG. 4/4/18. University of Salford.
2. Dos'Santos, T., Comfort, P., & Jones, P A. (2018). Between session reliability of lower-limb biomechanics during cutting: impact of trial size. BASES BIG. 4/4/18. University of Salford.
3. McBurnie, A. Dos'Santos, T., & Jones, P A. (2018). The performance-injury conflict during cutting. BASES BIG. 4/4/18. University of Salford.
4. Dos'Santos, T., McBurnie, A, Donelon, T., Thomas, C., Comfort, P., & Jones, P A. (2019). The Cutting Movement Assessment Score (CMAS): A Qualitative Screening Tool To Identify Athletes With High Peak Knee Abduction Moments During Cutting. BASES BIG. 10/4/19. University of Huddersfield.
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6. Dos'Santos, T., Comfort, P., & Jones, P A. (2019). Biomechanical effects of change of direction technique modification: implications for performance and injury risk. Salford Strength Symposium. 19/6/19. University of Salford.

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